Neuromuscular control of heel strike in running and walking: Does footwear midsole stiffness play a role?

A PCA approach to analyzing EMG waveforms

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Master thesis in Human Movement Science
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Trondheim, June 2015

Abstract

The current study had two main purposes. The first aim was to determine whether the findings from a previous study that found waveforms that could be related to mechanisms for feedback and feed-forward muscle activation in walking, by applying Principal Component Analysis (PCA) on the Electromyography (EMG) signals from muscles controlling the knee, could be reproduced in a second independent dataset. Furthermore, it was investigated if similar waveforms also could be found in muscles controlling the ankle joint, and that they would play a role in controlling the heel strike not only in walking but also in running.

The second aim was to get a better understanding of how the neuromuscular system adapts to specific heel strike conditions. This was done by systematical change the cushioning properties of a running shoe. It was hypothesized that the different shoe conditions would affect to what extend the characteristic feedback or feed-forward patterns would contribute to the muscle activation amplitude.

The study was conducted by having ten subjects walk and run with five different running shoes with different cushioning properties. EMG signals from muscles controlling the knee and ankle joint were collected while the subjects were walking and running. A PCA was conducted on the collected EMG signals, providing Principal Components (PCs) showing characteristic waveforms. Furthermore the effect of each shoe on these waveforms was found by statistical analysis.

The results from this study showed that waveforms that can be related to feedback and feed-forward muscle activation were found in the knee extensors both for running and walking. Waveforms that can be related to patterns of feedback and feed-forward muscle activation were not found for the knee flexors or for muscles controlling the ankle joint, neither for walking or running. Regarding the shoe effect, there were found no shoe effect on a group level that contributed to a systematic change in the waveforms. When looking for a shoe effect on an individual level the results suggests suggest that in walking, subject-specific adaptation to shoe sole stiffness seemed to take place in a significant fraction of the subjects. Indicating that there is an adaptation to shoes with different cushioning properties, but this adaptation takes place on an individual level and not in a common way for everyone. For running this individual adaptation was not seen in the same degree.

Acknowledgement

First I would like to thank my supervisor Professor Peter A. Federolf and co-supervisor Beatrix Vereijken for their help and guidance. I also want to show my gratitude to all the members of our research team, my fellow students Ådne Marlon Guldbrandsen and Lars Peder Dunfjell, and exchange student Mirko Kaufmann.
Furthermore I want to thank Tobias Goihl, Per Bendik Wik og Xiang chun Tan for their help in the lab and with the data collecting.
Last I want to thank the persons who volunteered to participate in this study.

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1 Introduction

Footwear is something everyone has a relation to, and most of us use it every day. We use shoes both as part of our daily living and as essential equipment for our recreational or sporting activities.

Research on footwear and its relation and impact to the human body has been going on for decades. Especially the moment of heel strike has been studied. Impact forces that occur when the heel collides with the ground at heel strike are large and the rate of force development is high. These forces have been associated with the development of injuries (Creaby & Dixon, 2008; Grimston, Engsberg, Kloiber, & Hanley, 2010; Lieberman et al., 2010; Milner, Ferber, Pollard, Hamill, & Davis, 2006; Zifchock, Davis, & Hamill, 2006).There have been several strategies for trying to control and reduce these forces. One of the most common strategies has been to make some kind of cushioning by changing the hardness of the shoes' midsole. However, studies examining the effect of this so far have been inconclusive. Some studies have found that a softer shoe decreases the impact force, while others have found that the impact forces increase with a softer shoe (Benno Maurus Nigg, 2010).In part these discrepancies might be explained with mechanical models (Shorten & Mientjes, 2011), but neuromuscular adaptation is also believed to play an important role. However, how the neuromuscular system adapts to shoes with different cushioning properties is not well understood. One hypothesis might be that the neuromuscular system regulates leg stiffness, e.g. by adjusting co-activation of antagonistic muscles. For example, a more rigid leg at heel strike would imply that the impact force peaks are higher. One speculative hypothesis might then be that the neuro-muscular system adjusts to different cushioning properties of shoes by regulating the leg stiffness such that the impact force is controlled. Another hypothesis might relate to the concept of "muscle tuning" (B. M. Nigg & Wakeling, 2001), which suggests that neuromuscular adaptation is regulated with the aim of controlling vibrational responses of the soft tissue compartments after impact.

An effective tool to study neuromuscular control is Electromyography (EMG). The musculoskeletal system has a high step-to-step and inter-subject variability in muscle activation patterns (Araújo, Duarte, & Amadio, 2000; Nair, French, Laroche, & Thomas, 2010; Winter & Yack, 1987). Even though EMG signals are highly individual, rhythmicity and timing of muscle activation are comparable between individuals (Bizzi, Cheung, d'Avella, Saltiel, & Tresch, 2008; Guidetti, Rivellini, & Figura, 1996; Huber, Nüesch, Göpfert, Cattin, & von Tscharner, 2011). The use of principal component analysis (PCA) on EMG signals has proved to be useful. PCA is well suited for finding patterns that are otherwise difficult to observe, and is therefore well suited to use on the highly variable EMG signals. PCA has made it possible to extract information from the EMG signal describing neuromuscular processes. By applying PCA on the EMG signals the main features of the EMG patterns could be described by relatively few underlying components.

Huber et al. (Huber et al., 2013) conducted a study where they investigated potential sources of intra- and inter-subject variability in the activation patterns of muscles stabilizing the knee joint 200ms before and after heel strike in barefoot walking by using PCA. Differences in heel strike characteristics are likely to be an important source of EMG variability making PCA a suitable tool for investigating adaptations to heel strikes. Huber and colleagues argue that the neuromuscular system has two general pathways to adjusting to specific heel strike conditions: a feed-forward strategy and a feedback-strategy. For each strategy, Huber et al. predicted characteristic waveforms and hypothesized that a PCA conducted on the EMG signals of muscles involved in controlling heel strike would reveal these characteristic waveforms as principal component (PC)-vectors. The first waveform, which was related to feedback mechanisms, would have a peak 30-50 ms after heel strike. The 30-50 ms would correspond to the reaction times of involved reflex circles (Figure 1a). The second characteristic waveform, which would be associated with pre-activation of the muscle before heel strike (feed-forward mechanism) was expected to show activation before heel strike and up to 30-50 ms after, followed by a sharp decline when the feed-back mechanism allows for adjusting to the specific heel strike conditions that took place (Figure 1b). Huber et al. conducted measurements of

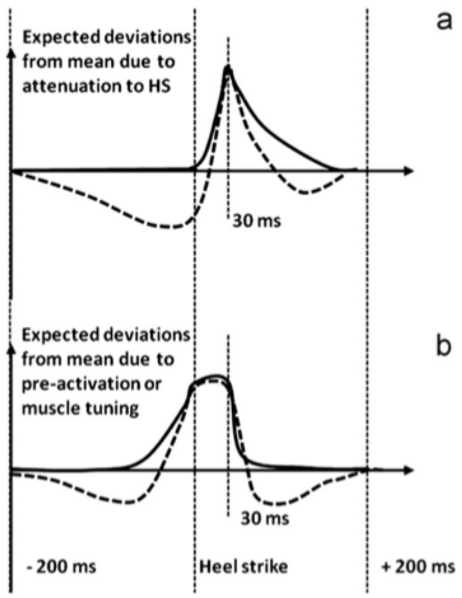

Figure 1 Predicted components of the EMG waveforme derived from conceptual considerations that may play a role in the adaptation to the heel strike (HS)(Huber et al., 2013)

five muscles controlling the knee: mm. rectus femoris, vastus medialis, vastus lateralis, semitendinosius and biceps femoris. They found that the shape of the first two PCs agreed well with the predicted waveforms of these muscles. Furthermore, would it be interesting to see if the same findings could be found not only for walking but also for running. It would also be interesting to see if the same applies to muscles that control the ankle joint in addition to the knee joint. And if the same applies if the heel strikes characteristics were changed.

The current study had two main purposes. The first aim was to determine whether the findings of Huber et al. can be reproduced in a second independent dataset. Furthermore, the study tested the hypotheses that similar waveforms may also be found in muscles controlling the ankle joint, and that they would play a role in controlling the heel strike not only in walking but also in running.

The second main aim was to better understand the mechanism of how the neuromuscular system adapts to specific heel strike conditions. Thereto the mechanical conditions at heel strike were artificially altered by having participants walk or run with shoes that had different cushioning properties. It was hypothesized that the different shoe conditions would affect to what extend the characteristic feed-back or feed-forward patterns would contribute to the muscle activation amplitude.

2 Method

2.1 Participants

A total of 10 subjects were included in the study. Their mean age (SD, min – max) was 25.9 years (±5.02, 21-37), weight was 73.43 kg (±6.79, 59.8 – 84.8), height was 178.45 cm (±4.83, 170.5 – 185.5) and BMI was 23.01 (±1.28, 19.98 – 24.64). Participants were recruited by seeking volunteers in various groups of local runners on Facebook and among people who exercise at the local athletics training facilities. To be eligible to participate, they needed to be between 18-40 years old, be healthy, have no recent (last 2 years) injury of the lower extremity, no history of any injury that could affect walking and running patterns, and they should be heel-toe walkers and heel-toe runners. In addition, they needed to have shoe size 42 (EUR) because the shoes being tested were custom made and were available only in this size. To secure that all participants had the same dominant foot, they were asked when showing interest to participate if they were left- or right-footed. Since the vast majority reported to be right-footed they were included, while those who reported to be left-footed were excluded. While an effort was made to find subjects of both sexes, it was not possible to recruit a sufficiently large group of females with the compulsory shoe size. Therefore, the current study includes only data from male volunteers.

The study was evaluated and approved by the regional medical ethical committee prior to participant inclusion, and all subjects signed an informed consent prior to participating in the study.

2.2 Shoes

Five different shoes were tested. The shoes were donated by Li-Ning, China, and were especially built for research purposes. The shoes were almost similar except for the midsoles that had different mechanical properties. Mechanical properties for the different shoes are shown in Table 1 and Table

Figure 2 The shoe that were tested

2. The differences between the shoes were mainly that the heel part and the front part of the soles had different stiffness across the five different shoes.

Table 1 Cushioning properties of the different shoes (Data provided by Thorsten Sterzing, Li-Ning)

Shoe condition			**Harndess**		
Shoe	*Foot*	*Called*	*Rearfoot*	*Midfoot*	*Forefoot*
				Durometer points	
Very very soft	Left	A	33.6	51.0	35.2
Very very soft	Right	A	35.6	51.4	24.4
Very soft	Left	B	43.0	52.4	43.4
Very soft	Right	B	43.0	53.2	43.2
Soft	Left	C	49.2	52.4	49.8
Soft	Right	C	48.2	54.0	48.8
Medium	Left	D	56.0	56.0	55.2
Medium	Right	D	55.0	55.2	56.0
Hard	Left	E	61.8	52.2	62.2
Hard	Right	E	62.6	51.0	63.6

Table 2 Mechanical data about the different shoes (Data provided by Thorsten Sterzing, Li-Ning

Shoe condition			**Midsole**		**Shoe**	
Target	*Foot*	*Called*	*Weight (g)*	*Length (mm)*	*Thickness (mm)*	*Weight (g)*
Very very soft	Left	A	159	285.0	25.0	328
Very very soft	Right	A	157	285.0	25.0	327
Very soft	Left	B	177	285.0	25.0	343
Very soft	Right	B	171	285.0	25.0	338
Soft	Left	C	181	285.0	25.0	349
Soft	Right	C	177	285.0	25.0	344
Medium	Left	D	198	285.0	25.0	367
Medium	Right	D	195	285.0	25.0	363
Hard	Left	E	201	285.0	25.0	368
Hard	Right	E	198	285.0	25.0	365

2.3 Measurement procedures

All subjects came to the lab for one session that consisted of both walking and running with the different shoes. Before the tests started, height and weight was measured for every subject, and the subject signed a consent form.

Thereafter the subjects changed clothes to a suitable outfit; they only wore short tights. The skin was then prepared for the EMG measurements, and EMG electrodes, and retro-reflective markers were placed at the correct position on the body of the subjects. (Details of this are described in the next section).

The subjects then walked and ran with five running shoes with different cushioning properties, and one bare foot trial. Starting with the walking trials followed of the running trials fore each condition. The order of the shoes was randomized to avoid that the shoes being worn in a specific order could affect the results. The bare foot trial was always the last trial.

The subjects walked and ran along a 10m walkway. They walked at a self-selected speed with the instruction to walk at a pace they would use if they were trying to catch the bus. Across trials, they could not deviate from the selected speed with more than +/- 10%. For the running trials, the subjects were instructed to use a speed of 3.3 m/s. As with the walking trials, the speed could not differ with more than +/- 10% across running trials. To control the speed a wireless timing system (TC Timing System, Brower Timing System, USA) was used. Photo gates were put up at a distance of 3.30m, covering the area over the force plates, and the time for every passing was recorded. If one passage was outside the permitted time range relative to the speed they had chosen for the walking trials or the speed they were instructed to use for the running trials, the recorded data for that passage was excluded from further data analysis.

The number of times the subjects walked and ran the 10m walkway for each shoe varied, depending on how fast they got enough good hits on one of the force plates mounted in the floor. To be a good hit the right foot needed to be inside the border of one of the force plates during an entire stance phase. There were three force plates mounted in the floor. Three trials with at least one clean hit were considered to be enough. The subjects were not instructed to try to hit the force plates in order to avoid that they would adjust their gait aiming for the force plates, because that would affect the way they walk or run. The number of actual trials varied between shoe conditions, since for some trials the subject could get three good hits quickly, while for others they needed many passes over the force plates to get the required number of hits.

Between each shoe condition the subjects answered some questions about the shoes on a questionnaire. Details about the questionnaire are described in an own section.

2.4 Instrumentation

The study protocol measured thigh and calf muscle activity with EMG of the right leg during leveled walking and running. Surface EMG signals were recorded (Noraxon U.S.A. Inc., Scottsdale, AZ, USA) with from 8 muscles in the right leg: mm. vastus lateralis, rectus femoris, vastus medialis, biceps femoris, semitendinosus, gastrocnemius, soleus and tibialis anterior. The pregeled, bipolar Ag/AgCL surface electrodes with an inter-electrode spacing of 17.5mm (Noraxon U.S.A. Inc., Scottsdale, AZ, USA) were placed on the muscles according to the SENIAM guidelines (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000), see Figure 3 for illustration. The EMG data were sampled at 1500 Hz.

Simultaneous three-dimensional movement data was collected using 3D motion capture recording system for human movement (Vicon, Oxford, UK). Cameras, capturing at 250Hz, recorded the position of retro-reflective markers placed at selected places on the legs and feet. Reflective markers were placed on spina iliaca anterior superior, spina iliaca posterior superior, trochanter major, lateral on the lower 1/3 of the tigh, lateral on the knee joint axis, lateral on the lower 1/3 of the lower leg, lateral malleolus of the ankle, anterior on calcaneus, and on the head of the 1. metatarsus, on both legs and feet, see Figure 3 for illustration.

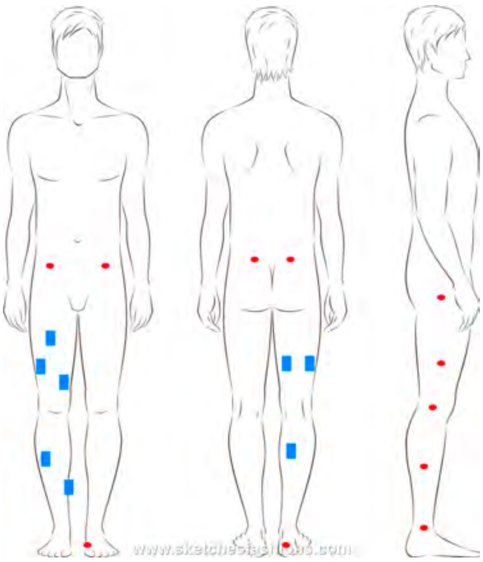

Figure 3 Illustration of the placement of the reflective markers and the EMG sensors. Red dots are the reflective markers and blue rectangles are the EMG sensors.

Ground reaction forces (GRF) between the foot and the ground were measured as the subject walked and ran over force plates (AMTI, Advanced Mechanical Technology, Inc., MA, USA) mounted into the floor. Forces were measured at 1500Hz.

One accelerometer (Noraxon U.S.A. Inc., Scottsdae, AZ, USA) was placed on the heel of the right shoe, and on the calcaneus on the right foot for the bare foot trials. This was used to calculate at what time the heel strike occurred.

2.5 Questionnaire

All subjects were asked to give subjective feedback on the shoes after both the walking and running session. They were asked to rate the shoes on a Visual Analogue Scale (VAS) (see Appendix) from very uncomfortable to very comfortable. In addition, they were asked to rate the shoes using a VAS from very soft to very hard. The subjects were also asked if they felt that the last shoe they used was more or less comfortable compared to the previous shoe, and if they felt it was softer or harder than the previous shoe. Lastly, they were asked to say something about the reason why they felt the shoe was comfortable or uncomfortable.

2.6 Data processing and statistics

All channels from all trials were visually inspected to identify and exclude from further analysis the trials or channels contaminated with spikes or other artefacts. The force plate signal was used to determine the steps when the volunteer hit the force plate. To obtain more data for the EMG analysis, not only "perfect hits" were included but also trials where the stance phase was distributed over two force plates. Thus, for each condition 10 step cycles could be extracted. The beginning and end of each step cycle was defined as the moment of the heel strike of the instrumented leg. A matlab-code was developed to automatically detect the moment of heel strike from the acceleration signal. However, to ensure that no falsely identified heel strikes could affect the analysis, the waveform of the acceleration signal of all individual steps was again visually inspected. Since in most subjects the acceleration waveforms differed substantially between barefoot and shod conditions, it was decided not to include barefoot in the following analysis.

The EMG signals of the selected steps were analyzed with a time-frequency analysis using 11 non-linearly scaled wavelets (von Tscharner, 2000), with center frequencies from 37 to 395 Hz. This frequency range was chosen since EMG movement artifacts typically affect frequencies lower than 30 Hz (Conforto, D'Alessio, & Pignatelli, 1999) and since the EMG spectrum typically has negligible power in frequencies higher than 400 Hz . The intensity of the EMG signal was defined as the sum of the intensities extracted from each wavelet with center frequencies from 37 to 395 Hz. EMG intensity was analyzed from 200ms before heel strike to 200ms after heel strike. This was called a waveform. Since the sampling frequency was 1500 Hz, 600 data points represented each waveform. The 300th time frame in each waveform was the heel strike as determined from the accelerometer data.

A normalization had to be developed that, on the one hand, makes possible the comparison of waveforms between subjects, and on the other hand preserves effects of the shoe conditions. As a first step, the power of each waveform (intensity integrated over the 400 ms) was calculated. Then the mean power of each subject's 50 trials (5 shoes and 10 heel strikes per shoe) was calculated and all waveforms of the subject normalized to this subject-specific average. For each muscle, the normalized waveforms of all subjects and all shoe conditions were then assembled into a matrix that was submitted to a PCA. The dimension of the input matrices was either 450x600 or 500x600 since all data of a subject was removed if artifact-contaminated channels were found in one or several trials of that subject.

A PCA was conducted separately of each muscle and separately for the walking and running trials. Each PCA yielded (Daffertshofer, Lamoth, Meijer, & Beek, 2004): (1) eigenvectors (PC-vectors) whose shape characterized correlated changes from the overall mean waveform; (2) eigenvalues, which quantify how much of the entire variability between waveforms was represented by each associated PC-vector; (3) "scores", obtained from projecting the individual waveforms onto the PC-vector, i.e. the scores quantify how much the specific pattern quantified each PC-vector contributed to the waveform observed in a specific trial.

The first goal of the current study was to provide further evidence for the hypothesis (Huber et al., 2013) at some PC-vectors obtained in a PCA on EMG waveforms around heel strike represent contributions of feed-forward and feed-back mechanisms for the control of the heel-strike event. Therefore, the first step in the analysis of the current data was to visually assess if the PC-vector waveforms agreed with the shapes that Huber and colleagues had predicted. Specifically, it was assessed, if the PC-vectors exhibited a peak at approximately 40 ms post heel strike (feedback mechanism) or a plateau between heel strike and 40ms post heel strike followed by a sharp decline (feed-forward mechanism).

The second goal of the current study was to determine if sole stiffness might affect the muscle activation patterns, and specifically the feed-forward or feedback components of muscle activation. Thereto two types of statistical analysis were conducted. First, it was assessed if group effects existed between shoe conditions. Hence, the 10 scores for each subject's trials in one shoe condition were averaged.

Then a repeated measures ANOVA was conducted on the averaged scores to determine a potential inter-subject shoe effect. Second, a intra-subject analysis was conducted by calculating an ANOVAs on the 50 scores obtained for each subject to determine if adaptation to the shoe condition might take place within some individual subjects. The threshold for statistical significance was set to a = 0.05. All analyses were conducted in Matlab (R2014b, The MathWorks Inc., Natick, MA, USA).

Note: Given the aim and the time scale of this thesis, and the complicated EMG analyzes. It was not prioritized to analyzing marker data that was collected during the trials.

3. Results

3.1 Waveform characteristics and variability of muscle activation during heel strike

A visual representation of the observed waveforms and their variability for the different muscles collected for all subjects and all shoe conditions are shown for walking in Figure 4 and for running in Fig. 2. In walking, the knee extensors show activation that gradually starts from around 100ms before heel strike and builds up to

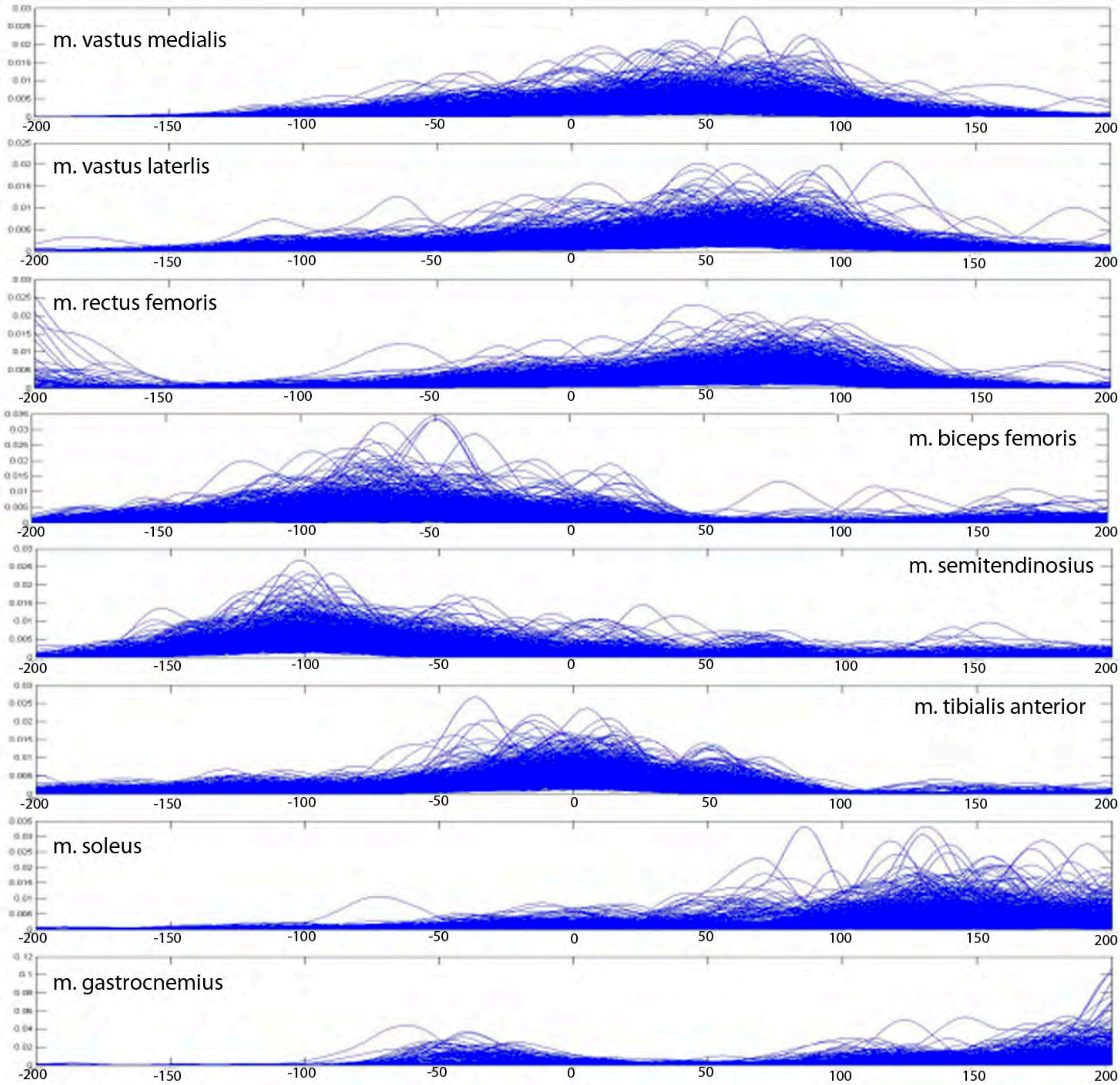

Figure 4 Waveforms for all subjects walking with the five different shoe conditions for the eight different muscles. Time 0 in the timeline on the x-axis shows the heel strike, and the figure shows the waveform 200ms before and after heel strike. The amplitude of the waveforms was normalized as described in the methods section.

around 60-70 ms after heel strike where they are most active. Thereafter the activation decreases more quickly and has very little activity from 100ms after heel strike.

For the knee flexor the activity in walking builds up from around 100 ms before heel

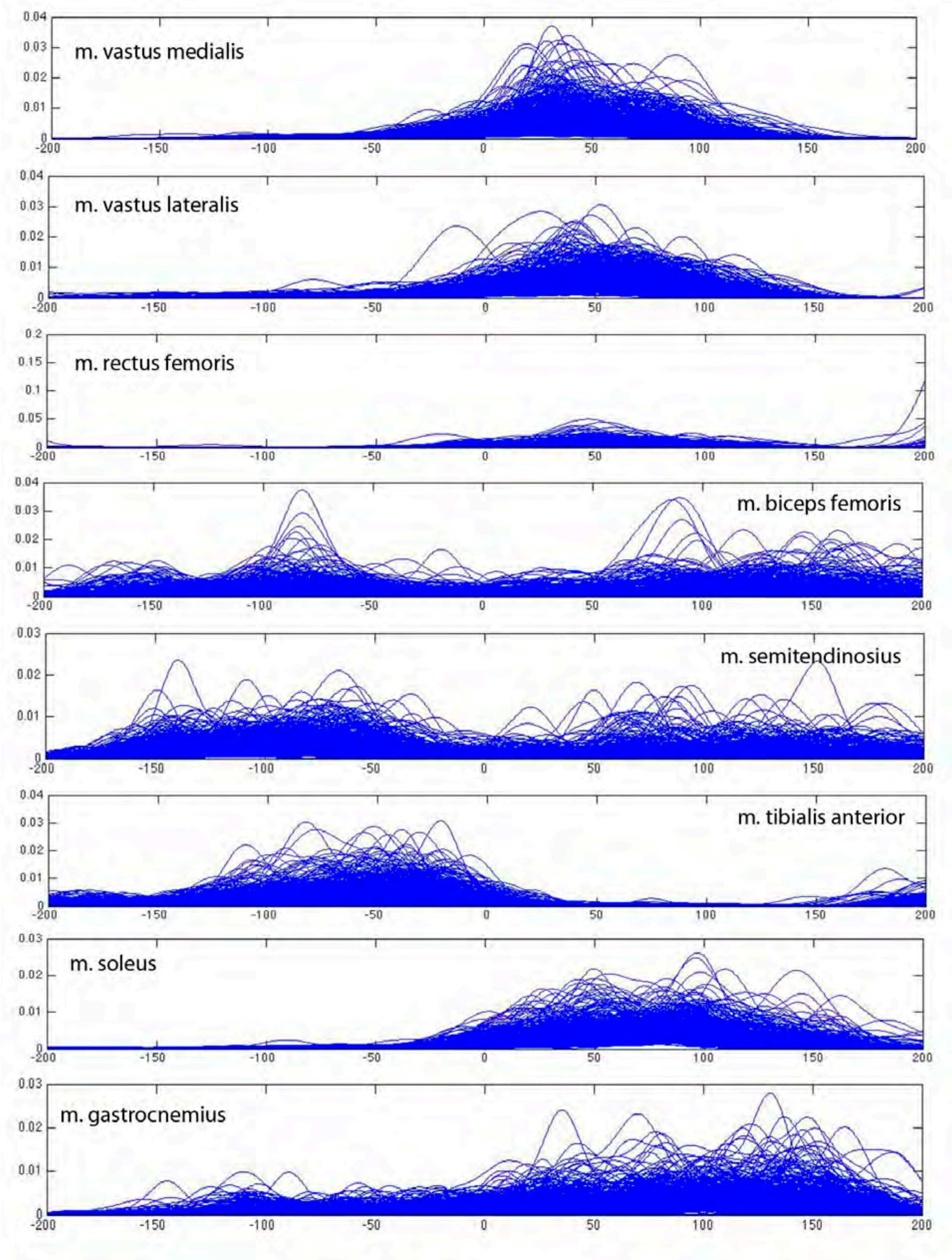

Figure 5 Waveforms for all subjects running with the five different shoe conditions for the eight different muscles. Time 0 in the timeline on the x-axis shows the heel strike, and the figure shows the waveform 200ms before and after heel strike. The amplitude of the waveforms was normalized as described in the methods section.

strike. With a peak in activity around 50 ms before heel strike, and gradually fades out from there to approximately 50 ms after heels strike. The ankle dorsiflexor, m. tibialis anterior, are in walking most active around heel strike. It builds up activity from around 100 ms before heel strike and the activity fades out to around 100 ms after heel strike. The activity of the plantar flexors builds up from right after heel strike and they are most active from 150 ms after heels strike and in the rest of the measured time period.

For running, the knee extensors are most active around 50 ms after heel strike. This is approximately around the same time period as in walking. But compared to walking the knee extensors are now active in a shorter time interval, from around 50 ms before heel strike to around 150 ms after heel strike. For the knee flexors it is more difficult to point out a time where they are most active in running, it seems like they are quite active through the whole time period measured. The ankle dorsiflexor is most active a little bitt earlier for running compared to walking. The activity peak is now around 50 ms before heel strike, building up from 150m ms before heel strike and fading out to around 50 ms after heel strike. For the ankle plantar flexors it also looks like they are active a bit earlier in running compared to in walking. The activity peaks are now around 100 ms after heel strike, building up from around heel strike and fading out to around 200 ms after heel strike.

A substantial proportion of the overall waveform was represented by only a few PCs. Across all eight muscles PC_1 – PC_4 accounted for 74% to 86% of the total waveform variability in walking and 62% to 93% of the total waveform variability in running. An overview of the first four PCs calculated from all waveforms from all subjects and all shoe conditions for the different muscles are shown in the Appendix (for walking in Fig. A1 and for running in Fig. A2).

3.2 PC-vector shapes related to feedback mechanisms in control of heel strike

Based on conceptual considerations and based on the article by Huber et al. it was hypothesized that the shapes of some PC-vectors would show distinctive features. The main feature associated with a feedback mechanism would be a peak approximately 30-50ms after heel strike. In walking, this feature was observed in the knee extensor muscles, specifically in mm. vastus medialis (Figure 6), vastus lateralis and rectus femoris in PC2. However, none of the PC-vectors in the knee flexors or in the muscles controlling the ankle joint exhibited this specific feature. In running, all three knee extensors, mm. vastus medialis, vastus lateralis and rectus femoris , showed this feature in the PC1 vector (Figure 3, right column). However, similar to walking, none of the PC-vectors in the knee flexors or in the muscles controlling the ankle joint exhibited this specific feature.

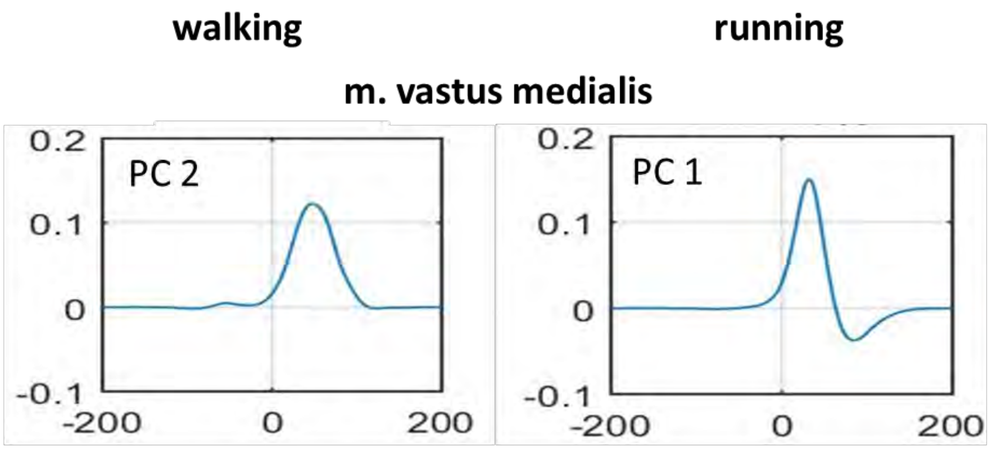

Figure 6 Shapes of the PC-vectors that showed a peak in the time range that might be associated with reflex mechanisms reacting to the heel strike (30-60ms after heelstrike).

3.3 PC-vector shapes related to feed-forward mechanisms in control of heel strike

A feed-forward adaptation was expected to produce a PC-vector shape with preactivation of the muscle before heel strike that peaks at or right after heel strike, followed by a sharp decline in activation at the time when the feedback controlled muscle activation peaks. In walking, such a characteristic was seen in PC1 for mm. vastus medialis (Figure 7), vastus lateralis and rectus femoris. And again, it was not seen in the knee flexors, mm. biceps femoris and semitendinosus, or in the muscles controlling the ankle joint, mm. tibialis anterior, soleus and gastrocnemius medialis. Qualitatively, the same observations were made for running in the PC2 vector (Figure 7).

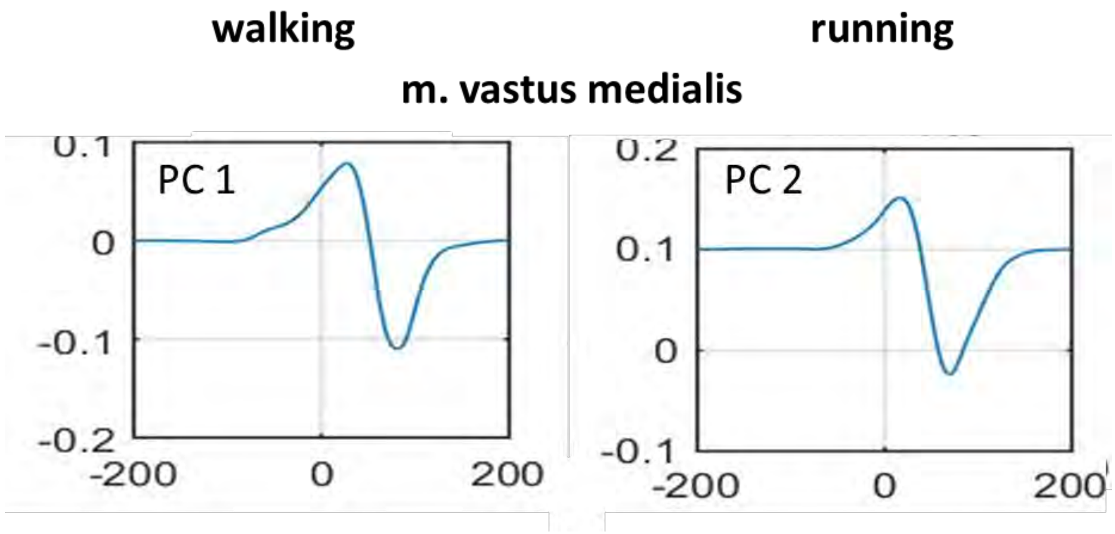

Figure 7 Shapes of the PC-vectors that showed a peak in the time range that might be associated with feed-forward mechanisms prior to the heel strike (30-60ms after heel strike.

3.4 Neuromuscular adaptation to changes in the cushioning properties of footwear

To determine whether there were any shoe effects in the PC scores, box plots were created. An overview of all boxplots created for the first 4 PC-scores of all eight muscles is included in the Appendix.

Overall the box plots did not show any systematic pattern that could correspond to the change of shoes and there were very few significant effects. For walking there was one significant effect, PC2 for m. soleus. But the waveform suggests that this is an activation of the muscle that happens some time after heel strike and has a shape that is likely not related to a feed-forward or feedback reaction. For running there were significant results in PC1 for mm. vastus medialis and tibialis anterior. For m. vastus medialis it was not possible to see a pattern in the box plots, hence while a shoe effect seemed to exist, it did not appear to cause a substantial change in the activation. For m. tibialis anterior the shape of the waveform did not suggest that the PC1 is related to heel strike. For running there were also significant results for PC3 of m. gastrocnemius medialis, and PC4 of mm. rectus femoris, biceps femoris and semitendinosus. However, again PC3 – PC4 were not likely to be related to feed forward and feedback muscle activation patterns.

When looking at an individual level, the results were slightly different. Table 1 shows an overview of the number of subjects for whom a significant shoe effect was found in walking for each PC and each muscle. Table 2 shows the same for running.

Table 3 The number of significant PCs on an individual level for each muscle in walking

	Artifact free subjects	PC_1	PC_2	PC_3	PC_4	PC_5	PC_6	PC_7	PC_8	PC_9	PC_{10}
M. vastus medialis	12	3	6	3	2	4	0	4	3	0	0
M. vastus laterlais	12	4	5	1	3	3	5	2	2	2	3
M. Rectus femoris	12	6	6	3	4	3	3	3	0	4	3
M. biceps femoris	12	4	6	2	1	2	2	1	3	1	1
M. semitendinosius	12	4	1	2	0	2	3	3	2	2	3
M. tibialis anterior	12	8	5	8	2	3	4	1	3	0	2
M. soleus	9	3	4	2	4	3	5	4	3	1	4
M. gastrocnemius	12	3	2	3	5	0	5	5	2	1	1

Table 4 The number of significant PCs on an individual level for each muscle in running

	Artifact free subjects	PC_1	PC_2	PC_3	PC_4	PC_5	PC_6	PC_7	PC_8	PC_9	PC_{10}
M. vastus medialis	10	1	0	0	0	1	1	0	0	0	2
M. vastus laterlais	10	0	1	1	0	2	1	1	0	2	1
M. Rectus femoris	9	2	1	3	3	2	0	1	0	1	2
M. biceps femoris	10	3	0	0	2	1	1	3	2	2	2
M. semitendinosius	10	1	1	1	2	0	3	1	3	0	1
M. tibialis anterior	9	3	1	2	2	3	0	3	0	1	1
M. soleus	9	2	0	3	2	2	2	1	1	1	1
M. gastrocnemius	10	1	1	0	2	1	2	1	2	0	1

These results suggest that in walking, subject-specific adaptation to shoe sole stiffness seemed to take place in a significant fraction of subjects. Since the shoe effect was not visible on a group level, this suggests that the different subjects adapted in a subject-specific way. In running, significant intra-subject shoe differences could be observed in only a small fraction of subjects. This suggests that the shoe effects that were observed on the inter-subject level in running might have been coincidental results that should not be over-interpreted.

3.5 Subjects' perception of shoe stiffness

After the subjects finished testing one shoe and before they started to test a new one, they were asked to answer some questions about how hard or soft they felt each shoe was, and how comfortably it was to walk and run with. Figure 8 shows the ratings of how soft or hard they felt the shoes were.

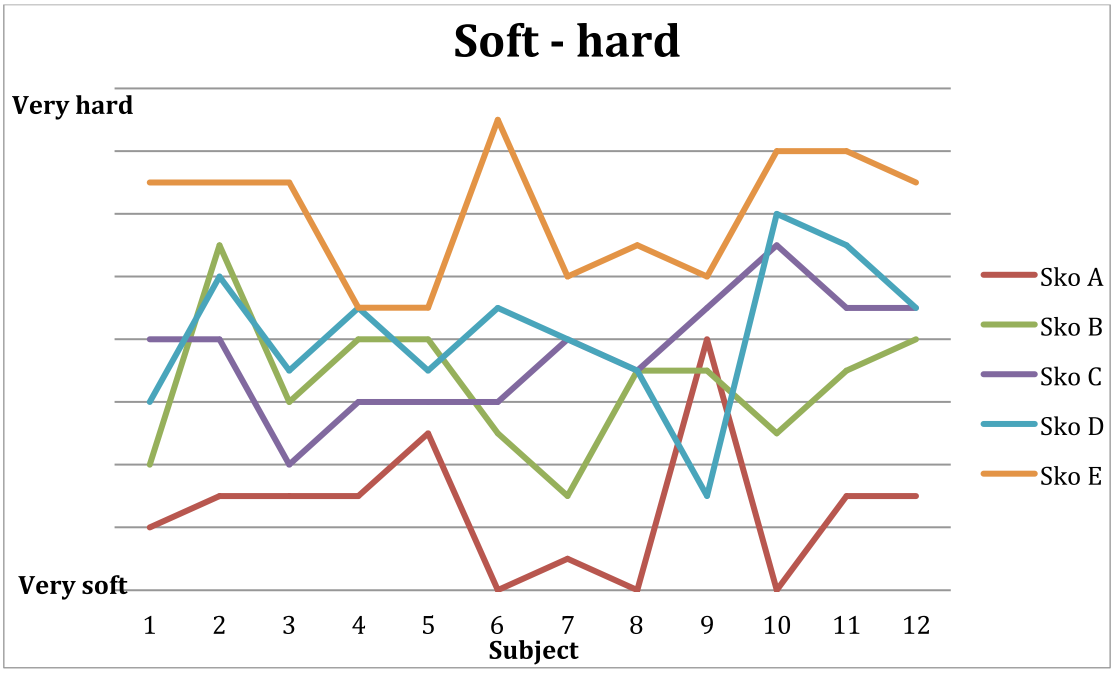

Figure 8 Ratings of the shoes by the different subjects from very soft to very hard using a VAS-scale. The different subjects are on the x-axis. The different shoes are shown with the colored lines, where shoe A was the softest one and shoe E the hardest. Subjects were connected for better identification of each shoe, not to suggest relations between the subjects.

Figure 8 shows that everyone was able to point out the hardest shoe and all but one could point out the softest shoe. For the three shoes with medium stiffness, the subjects found it more difficult to distinguish the shoes.

The different subjects also rated how comfortable they felt the shoes were to walk and run in from very comfortable to very uncomfortable using a VAS-scale. Figure 9 shows the ratings for walking and Figure 10 for running. Both figures indicate that the different subjects did not agree about what was the most comfortable shoe to walk and run in, despite that they could recognize the softest and the hardest shoe. The variation between subjects was substantial and it seemed to be highly individual what shoes the individual subjects preferred.

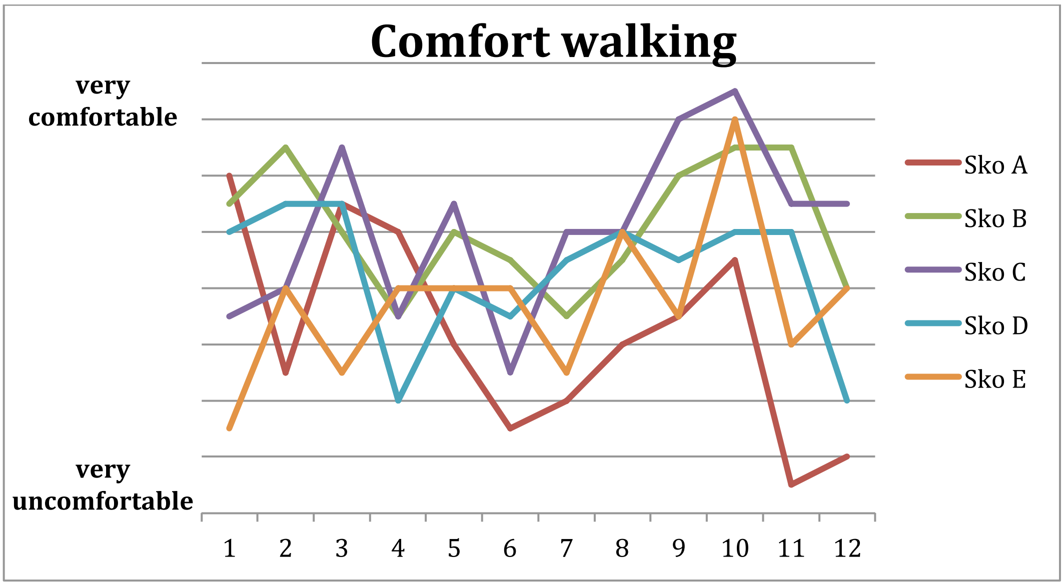

Figure 9 The different subjects rating on a VAS-scale of how comfortable they felt the different shoes were to walk in.. The different subjects are on the x-axis. The different shoes are shown with the colored lines. The different shoes are ranged from A the softest to E the hardest. Subjects are connected to help identify the different shoes, not to suggest relations between the subjects.

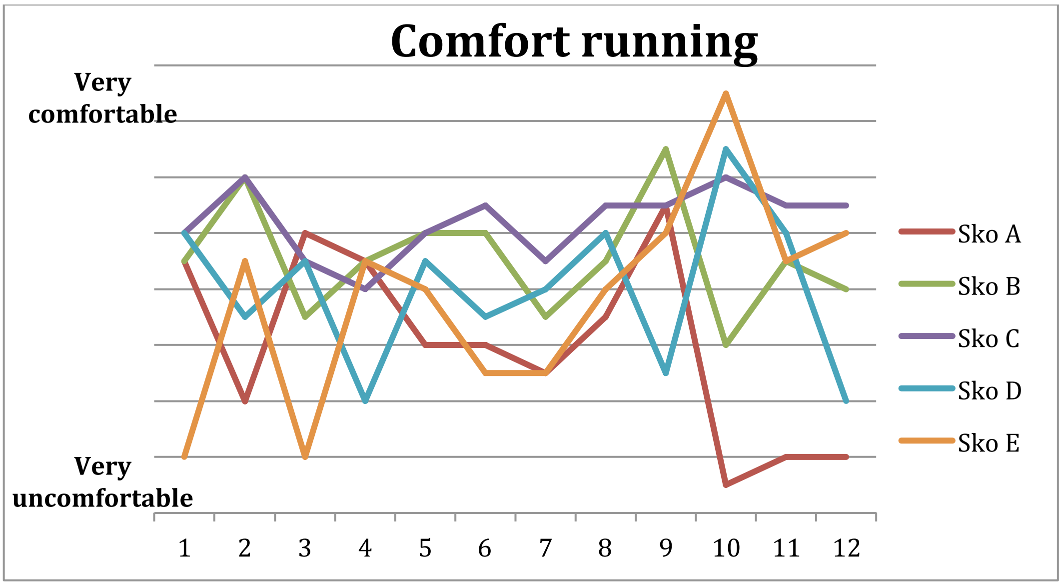

Figure 10 The different subjects rating on a VAS-scale of how comfortable they felt the different shoes were to run in. The different subjects are on the x-axis. The different shoes are shown with the colored lines. The different shoes are ranged from A the softest to E the hardest. Subjects are connected to help identify the different shoes, not to suggest relations between the subjects.

4. Discussion

The current study had two main purposes. The first aim was to see whether the findings of Huber et al. (2013) on identifying waveforms of the first PCs that could be related to feedback and feed forward muscle activity could be reproduced in a second independent dataset. Furthermore, this study tested the hypotheses that similar waveforms as identified by Huber in muscles controlling the knee joint may also be found in muscles controlling the ankle joint, and that they would play a role in controlling the heel strike not only in walking, as in Huber and colleagues had found, but also in running. The second main goal was to better understand the mechanism of how the neuromuscular system adapts to specific heel strike conditions. Thereto the mechanical conditions at heel strike were artificially altered by having participants walk or run with shoes that had different cushioning properties. It was hypothesized that the different shoe conditions would affect to what extend the characteristic feedback or feed forward patterns would contribute to the muscle activation amplitude.

4.1 Waveform characteristics

For three out of the five muscles Huber et al. studied, mm. vastus medialis, vastus lateralis and rectus femoris, we found a similar shape of the waveforms that might be associated with feedback and feed forward patterns of muscle activation. For these muscles, it is plausible to assume that adaptation to the heel strike event is a major source of variability. For the muscles mm. biceps femoris and semitendinosius the current study could not reproduce the same findings as Huber et al. The waveforms found here were different from the waveforms that were proposed by Huber et al. (2013) for feedback and feed forward muscle activation.

One reason that the results for mm. biceps femoris and semitendinosius were not congruent may be due to differences in the protocols. While Huber et al. (2013) tested the subjects barefoot, the current study tested them wearing footwear. Several studies (Bishop, Fiolkowski, Conrad, Brunt, & Horodyski, 2006; De Wit, De Clercq, & Aerts, 2000; Lieberman et al., 2010) have shown that there are differences in the biomechanics between barefoot runners and runners wearing shoes. Most importantly, a larger proportion of those running barefoot are likely to land flat or have a forefoot strike at ground contact, compared to those wearing shoes who mainly are rear-foot, heel striking at ground contact. This difference in landing between barefoot and wearing shoes is likely to be less visible when walking compared to running. One could speculate that the subjects who walked barefoot may have been less clear heel strikers compared to those who used shoes, and that this may have contributed to different results for these muscles.

A second aspect that could have contributed to that the results are not congruent for two of the five muscles studied could be that the studies are performed on different genders. Huber et al. used female subjects, while this study used male subjects. Ferber, McClay Davis, and Williams Iii (2003) found that female recreational runners had significantly different lower extremity mechanics compared to male runners. Although it is reasonable to assume that this difference might be smaller when walking than during running, it cannot be ruled out that gender differences might play a role.

Finally, a third aspect to consider is that the current study used a different normalization as compared to Huber et al. (2013): in Huber's study the normalization of the waveforms was done to unit intensity. In the current study differences between shoes were tried to be preserved by normalizing to a subject mean intensity. This difference in normalization is likely to have affected the specific shapes of the PC-vectors, however, it is unlikely the explanation for the differences observed in the knee flexor muscles between the current and Huber's study.

For running the current study found the same results as for walking. For the muscles mm. vastus medialis, vastus latearlis and rectus femoris the waveforms proposed and found by Huber et al. were reproduced. However, for the muscles mm. biceps femoris and semitendinosius we could not find waveforms that were similar to those proposed before. Here, as for walking, one could speculate that the reason for this might be due to the subjects being tested barefoot in Huber et al. compared to wearing shoes in the current study. Another potentially important difference is that the subjects went from walking to running. Several studies (Cavanagh & Kram, 1989; Dufek, Mercer, & Griffin, 2009; Mercer, Vance, Hreljac, & Hamill, 2002) have found that both the biomechanics and the impact forces change when the speed we move at changes. Both these changes, speed and shoes, are likely to influence the way a person contracts his muscles and may have contributed to the discrepancies between that the current study and Huber et al. (2013).

The current study also investigated whether the same shape in waveforms proposed for feedback and feed-forward muscle activation related to heel strike could be found in muscles controlling the ankle joint. Mm. tibialis anterior, soleus and gastrocnemius medialis were examined. However, neither for walking nor for running were waveforms found that resembled the proposed waveforms for feedback and feed-forward muscle activation. An explanation for this could be related to the time window of the gait cycle that was examined, 200ms before and after heel strike. Especially for the m. soleus and m. gastrocnemius medialis we can see from the visual presentation of the waveforms, Figure 4. and 5., hat they are most active around the border of the time window. This suggests that other neuromuscular processes, such as the beginning push-off, may have dominated the variation between waveforms in these muscles making phenomena related to heel strike more difficult to detect. Future studies looking at muscles controlling the ankle at heel strike should thus consider using a smaller time window to better focus on heel strike events.

4.2 Shoe effect

The second goal of this study was to better understand how the neuromuscular system adapts to specific heel strike condition. The participants walked and ran with shoes with different cushioning properties. We hypothesized that the different shoe conditions would affect to what extend the characteristic feedback or feed forward patterns would contribute to the muscle activation amplitude. Boxplots were crated showing the effect of each shoe on the PC scores. Overall the box plots did not show any systematic pattern that would correspond to the change of shoes. So on a group level it does not look like different shoes influences how the neuromuscular system reacts to the heel strike. In assessing shoe effect on muscles that control the ankle one should take into consideration that mm. soleus and gastrocnemius medialis were most active in the border of the time period examined. This may have led to that relevant data were not included in the analysis, and that an eventual shoe effect did not show

When we looks at the results on an individual level the results was a bit different. Here the amount of significant results was much higher, see Table 3. and 4. This indicates that there is a shoe effect on how the neuromuscular system adapts to the heel strike with the different shoes, but this adaptation is done differently for each individual. These findings may relate to theories about muscle tuning described by B.

M. Nigg (2001) and B. M. Nigg and Wakeling (2001). The main aspect of this theory is that the muscles are tuned before heel strike to dampen vibrations that occurs in the soft tissue package of the leg as a result of the collision of the heel with the ground. The shock wave made from the collision of the heel with the ground has a major frequency content of about 10-20 Hz(B. M. Nigg & Wakeling, 2001). The soft tissue package has natural frequencies between 5-65 Hz, depending on the activation, length, and contraction velocity of the muscles involved (Wakeling & Nigg, 2001). If these two frequencies coincide, it is thought that a resonance phenomenon can occur, which potentially might cause discomfort or in worst case be damaging for the part of the soft tissue package affected. To avoid this, the neuromuscular system can tune the muscles to avoid the frequency range of the shock wave crated from the heels collision with the ground. Muscle tuning is thought to be highly subject specific because vibration and damping properties of the soft tissue package depend on the mass and geometry of each individual(Boyer & Nigg, 2006; Benno M. Nigg & Liu, 1999; Pain & Challis, 2006).

There were more significant results for shoe effects in walking then for running. This could be, as described above, caused by that both the biomechanics and the impact forces changes when the velocity we move with changes. When a person is running, compared to walking, he is more likely to land more flat on the foot, i.e. the heel strike is not so prominent. This may lead to that the weight of the person are distributed on a slightly larger area, and the differences in cushioning properties between the shoes are of slightly less importance and not so noticeable.

4.3 Subjects' perception of the shoes

One issue that was considered at the start of the study was that the difference in the shoes cushioning properties could be too small to have any measurable effect on muscle adaptation. But if we look on Figure 8, showing how the different subjects rated the stiffness of the shoes, almost every one could point out what was the softest and what was the hardest shoe. This should indicate that the differences in cushioning properties between shoes were enough. Naturally, it was more difficult to distinguish the shoes that were more similar in stiffness from each other.

When we look on the Figures 9. and 10, showing how the subjects rated the comfort of the shoes, on can see that it is no consensus of what is the most comfortable shoe. This can fit well with the results showing that there was no common adaptation strategy on group level, but on an individual level there are individual adaptation strategies. There is not one shoe that fits all, but everyone have their individual taste.

4.4 Limitations

This study may also have some limitations. The population studied was quite small, 10 persons. Originally we wanted 16-20 people. The biggest issue recruiting subjects was to find subjects with the correct shoe size. On the other hand, PCA have previously been used on small populations and yielded good results. Therefore, the small population was not considered to be of a critical significant. Another thing that could be an issue concerning the populations was that it mainly consisted of trained athletes/runners. Some of them were very interested in running shoes and how the different shoes felt. So it may have led them to in a greater extent control the use of the shoes instead of letting a natural response to the neuromuscular system adapt to the shoes.

The size of the test lab could also be an issue to consider. The lab was not very long, meaning that the subject did not have much space outside the measuring area to accelerate and brake. For walking this should not be of any problem, but it may be for running. Since the running speed was very slow it was assumed that the subjects only needed 1-2 steps to reach the desired speed. The speed over the measuring area was also consistent for the subjects. Although one cannot exclude that the length played a role completely, it was considered that the length was sufficient to get good measurements.

4.5 Conclusion

To try to conclude, the current study found comparable waveforms for walking in the muscles responsible for knee extension as Huber et al. did in a previous study. This are waveforms that can be linked to feedback and feed-forward patterns of muscle activation. There were not found similar waveforms as Huber et al. (2013) did in muscles responsible for knee flexion. Further the current study examined if there could be found waveforms for the same muscles in running that were of the same shape as those proposed and found by Huber et al. indicating feedback and feed-forward patterns of muscle activation. As for walking comparable waveforms were found for the knee extensors, and not for the knee flexor.

In addition to the muscles controlling the knee, muscles controlling the ankle were investigated. Here neither was there found comparable waveforms indicating patterns of feedback and feed forward muscle activation.

Further more this study examined how the neuromuscular system adapted to shoes with different cushioning properties. In advanced, it was hypothesized that the different shoe conditions would affect to what extend the characteristic feedback or feed-forward patterns would contribute to the muscle activation amplitude. The results showed that there were no shoe effect on a group level on the waveforms that were characteristic for feedback and feed-forward patterns of muscle activation. When looking at an individual level, the results were slightly different. The results here suggest that in walking, subject-specific adaptation to shoe sole stiffness seemed to take place in a significant fraction of the subjects. Indicating that there is an adaptation to shoes with different cushioning properties, but this adaptation takes place on an individual level and not in a common way for everyone. In running there were only significant results for shoe adaptation in a small fraction of the subjects. Indicating that the shoe effect seen here might be coincidental and should not be over-interpreted.

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Appendix

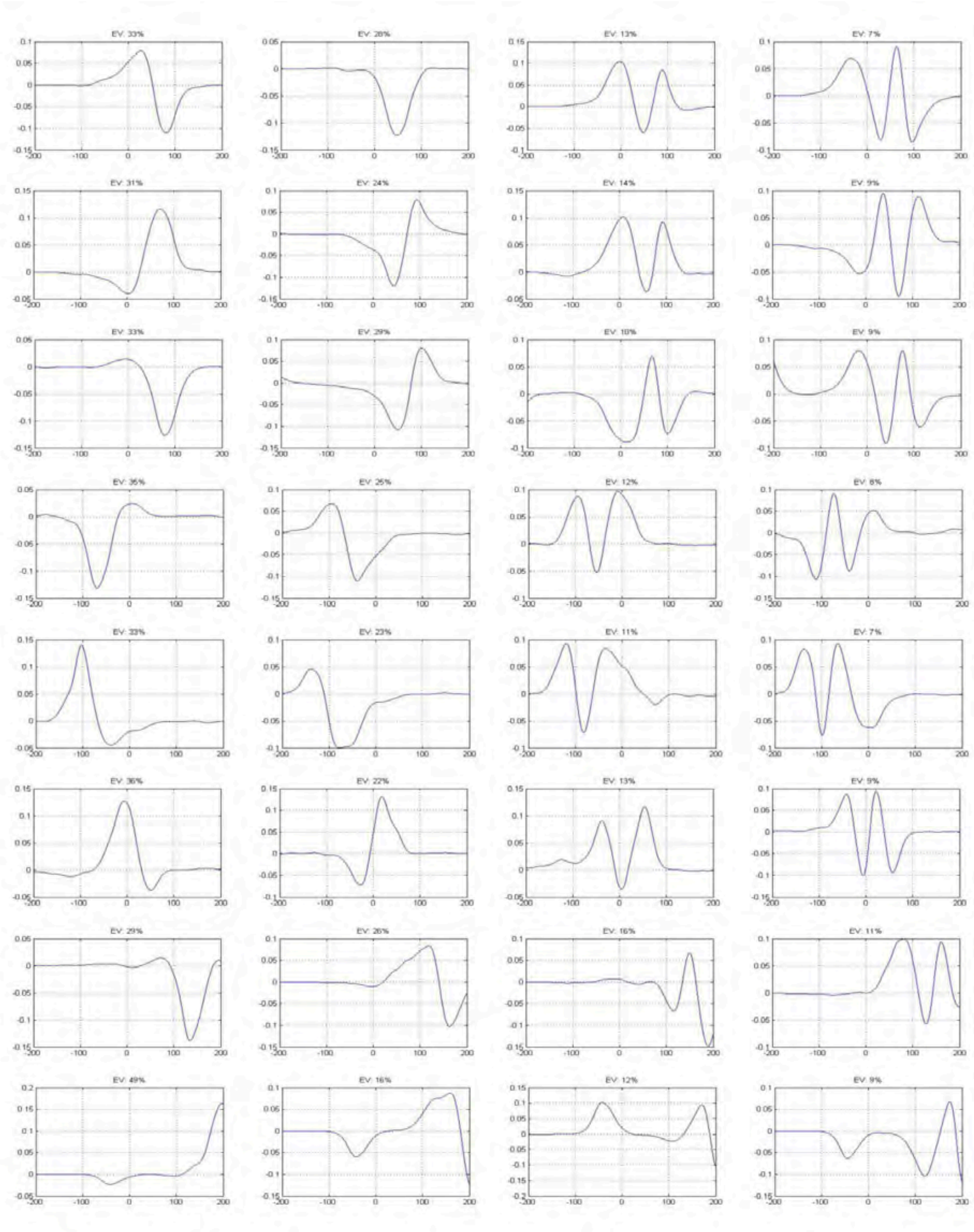


Figure A1 The four first PCs from the waveforms for all subjects and the five different shoe conditions for the eight different muscles measured in walking. From top to bottom: mm. vastus medialis, vastus lateralis, rectus femoris, biceps femoris, semitendionosius, tibilais anterior, soleus and gastrocnemius medialis.

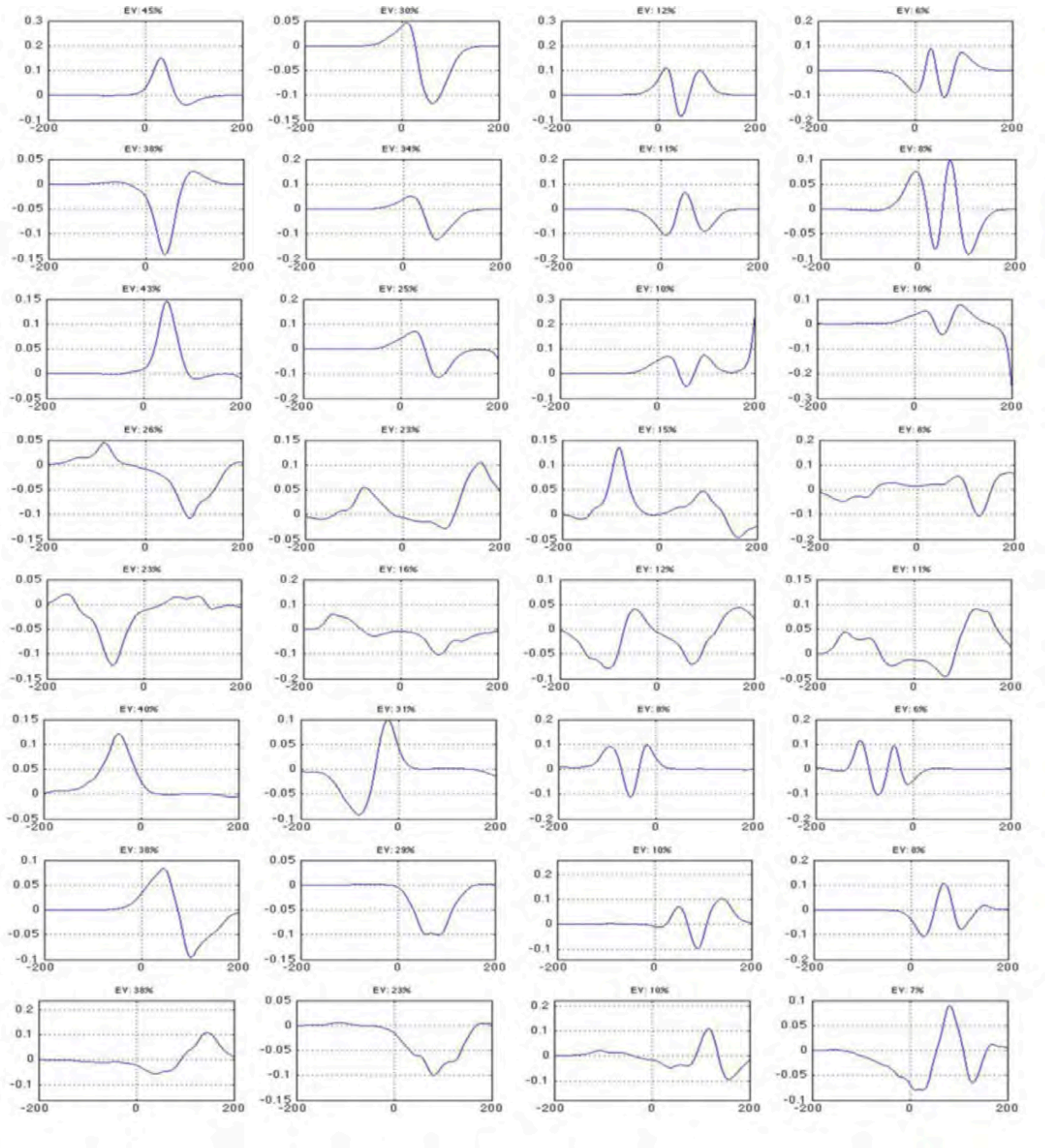


Figure A2 The first four PCs from the waveforms for all subjects and the five different shoe conditions for the eight different muscles measured in running. From top to bottom: mm. vastus medialis, vastus lateralis, rectus femoris, biceps femoris, semitendionosius, tibilais anterior, soleus and gastrocnemius medialis.

Note: It is important to remember that the direction of the PC-vectores is arbitrary. Hence, the shape or the inverse shape (mirrored at the y=0 line) need to be compeared to the predicted shape.

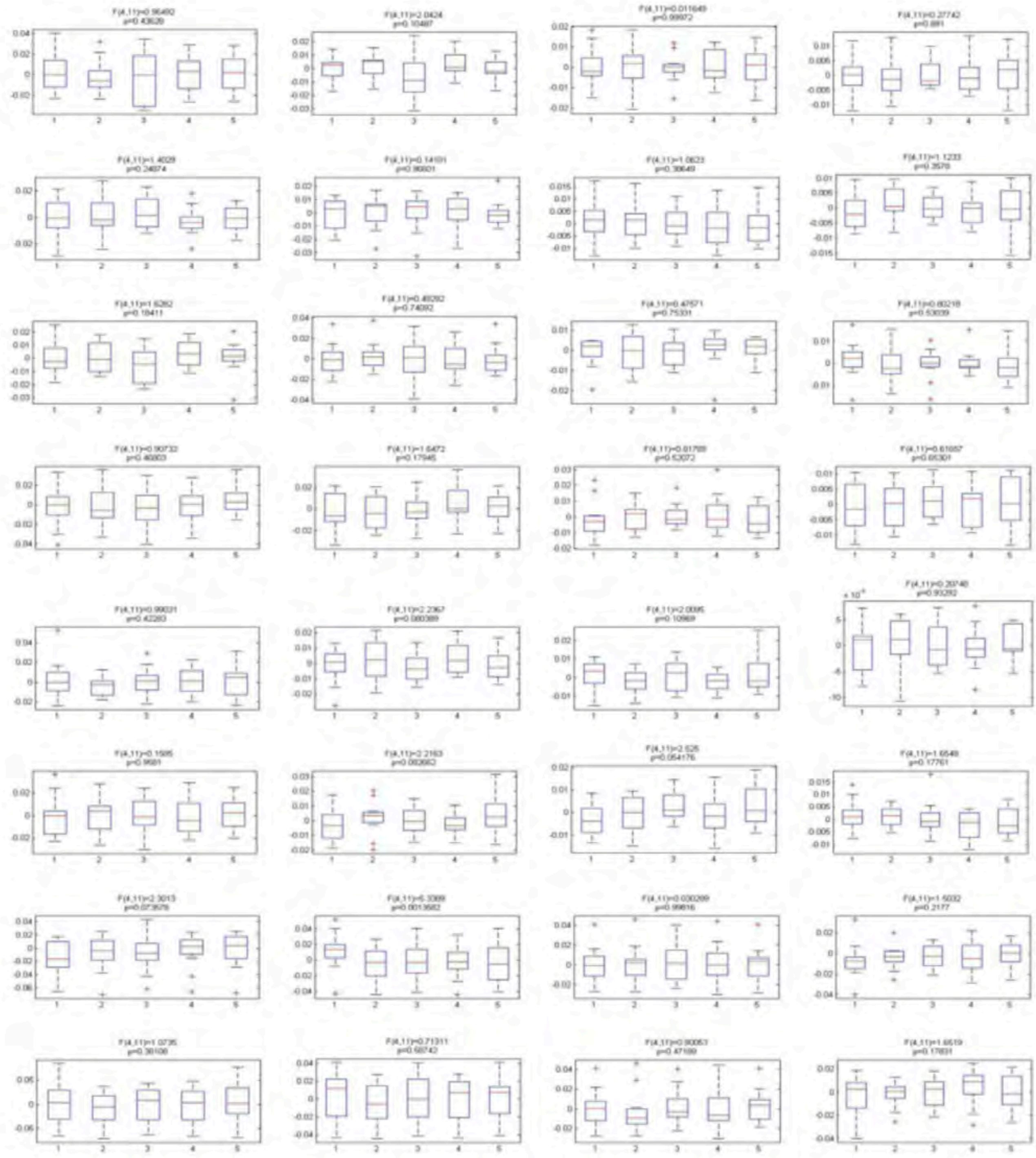


Figure A3 Boxplots showing if there were any shoe effects on the PCs for the eight different muscles in walking. From top to bottom: mm. vastus medialis, vastus lateralis, rectus femoris, biceps femoris, semitendionosius, tibilais anterior, soleus and gastrocnemius medialis.

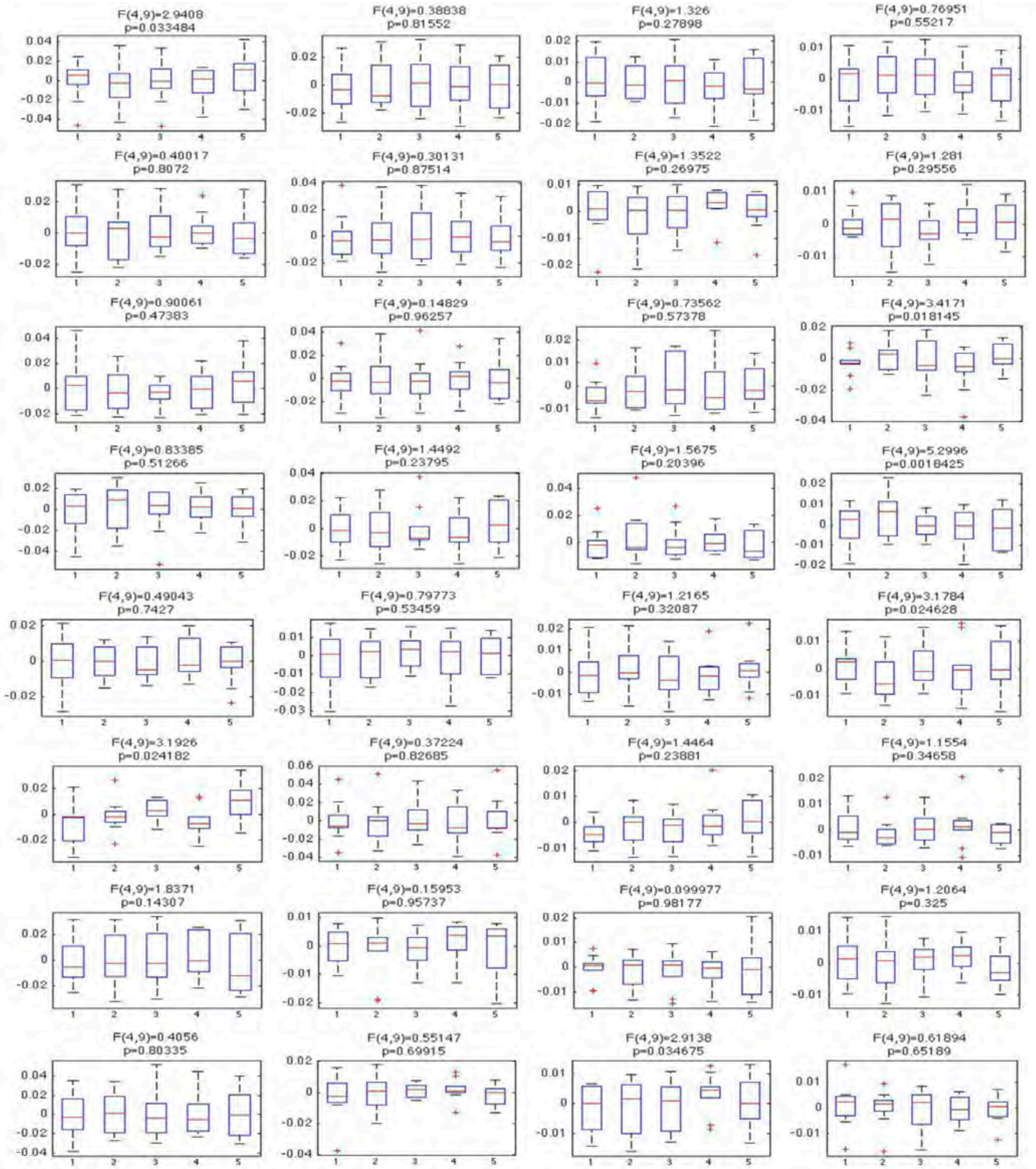
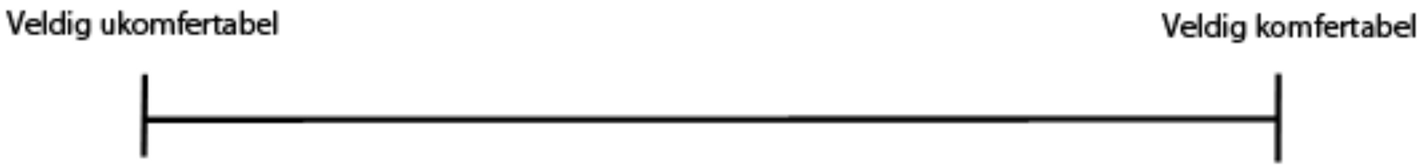


Figure A4 Boxplots showing if there were any shoe effects on the PCs for the eight different muscles in running. From top to bottom: mm. vastus medialis, vastus lateralis, rectus femoris, biceps femoris, semitendionosius, tibilais anterior, soleus and gastrocnemius medialis.

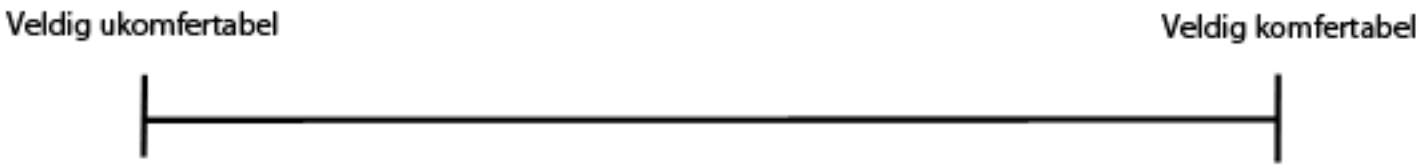
Spørreskjema

Sko 1:

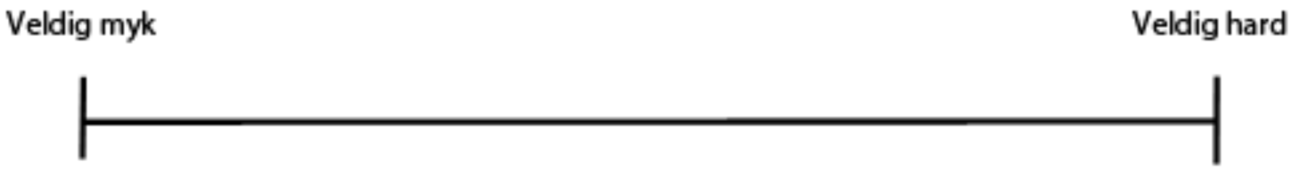
- Komfort ved gange



- Komfort ved løp



- Hvor hard syns du skoen føltes?



- Bare sammenlignet med den forrige skoen, er denne skoen:

- Mer komfortabel eller mindre komfertabel
- Hardere eller mykere

- Var det noen spesiell grunn til at skoen føltes komfertabel eller ukomfertabel?