Biomechanical and neuromuscular adaptations before and after realignment surgery for ankle osteoarthritis

Inauguraldissertation

zur
Erlangung der Würde eines Doktors der Philosophie
vorgelegt der
Medizinischen Fakultät
der Universität Basel

von
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aus Krattigen BE

Basel, 2013

Genehmigt von der Medizinischen Fakultät

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Basel, den 12. Juni 2013

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Summary

Osteoarthritis of the ankle joint develops mainly after trauma. It is less frequent than osteoarthritis of the knee or hip joint but often affects younger and physically active people. More than half of the patients present with a malalignment of the hindfoot, more frequently into varus. Such patients with asymmetric ankle osteoarthritis often have partially intact articular cartilage and may thus benefit from joint preserving realignment surgery. With supramalleolar and hindfoot osteotomies, the ankle is surgically realigned to unload the degenerated cartilage and improve the joint congruency. While clinical outcome studies already showed a reduction of pain, as well as improvements in the joint function and the patients' general well-being, the effects of supramalleolar osteotomies on the patients' gait patterns are largely unknown. The aim of this thesis was therefore to first identify and quantify the gait patterns of patients with asymmetric (varus or valgus) ankle osteoarthritis and secondly to assess the biomechanical and neuromuscular rehabilitation potential after joint preserving realignment surgery.

Chapters 2 and 3 focus on the characterization of the biomechanical and neuromuscular gait adaptations in patients with asymmetric ankle osteoarthritis. The results of the gait analyses in patients with early- to mid-stage asymmetric ankle osteoarthritis showed that these patients had a lower hindfoot dorsiflexion and rotation range of motion, as well as reduced peak ground reaction forces, external ankle dorsiflexion moment, and ankle joint power. These changes were similar to those previously described in patients with end-stage ankle osteoarthritis. Additionally, the application of a principal component analysis on the temporal waveforms of the hindfoot dorsiflexion angle and the vertical ground reaction force resolved features that influenced the amplitudes and timing of the waveforms. Using selected principal component scores of patients and healthy subjects in a linear support vector machine classifier resulted in a successful classification (recognition rate: > 95%). Hence, these results indicate distinct changes in the gait patterns of patients with asymmetric ankle osteoarthritis that can be resolved by principal component analysis (Chapter 2).

Neuromuscular adaptations in patients with ankle osteoarthritis include muscle weakness and spectral changes in the muscle activation. Chapter 3 showed that patients with asymmetric ankle osteoarthritis produce lower isometric torques in plantarflexion and dorsiflexion compared to healthy subjects. This weakness of the lower leg muscles was also associated with changes in the muscle activation patterns. For the *M. tibialis anterior* the wavelet power spectrum (maximal isometric contraction) and the wavelet pattern (walking) contained more low frequency components than those of healthy subjects. During walking, the calf muscles (*Mm. gastrocnemius medialis, gastrocnemius lateralis,* and *soleus*) were active with a lower intensity and over a broader time-frequency region. Additionally, the influence of the hindfoot alignment on the muscle activation was studied. Although that the number of subjects was small, it seemed that patients with a valgus hindfoot alignment have an altered intermuscular coordination between the calf muscles. While in healthy subjects and patients with a varus alignment *M. gastrocnemius medialis* was maximally active before *Mm. gastrocnemius lateralis* and *soleus,* it lagged behind in patients with valgus ankle osteoarthritis. This altered coordination could be due to a reduced or missing varisation of the hindfoot during push off from the floor. Further changes were seen in the wavelet patterns of *M. peroneus longus* that

contained more low frequency components in patients with a valgus hindfoot alignment than in those with a varus alignment. This could be related to a lower muscle activation level that has previously been described for healthy subjects with flat-arched feet.

Based on the described gait adaptations in patients with asymmetric ankle osteoarthritis, Chapters 4 and 5 illustrate the effects of the joint preserving realignment surgery and the following rehabilitation on the biomechanical and neuromuscular gait patterns. The first study presented in Chapter 4 investigated the long-term biomechanical outcome of supramalleolar osteotomies in a group of patients following at least seven years after surgery. For the spatiotemporal, kinematic, and kinetic gait parameters patients after realignment surgery showed fewer differences to controls than patients with asymmetric ankle osteoarthritis. The postoperative patients walked faster, with a higher cadence, and a slightly higher ankle dorsiflexion moment. However, the range of motion in hindfoot and hallux dorsiflexion remained reduced compared to healthy subjects. Additionally, prospective gait data for patients before and after realignment surgery were collected and presented together with data on the long-term follow-up patients in Chapter 5. For patients with ankle osteoarthritis, short-term (prospective) and long-term follow-up patients similar changes in the foot kinematics were seen. Principal component scores that affected the range of motion of the sagittal hindfoot and hallux movement were reduced compared to healthy subjects in all patient groups. For the forefoot dorsiflexion angle (range of motion) and the temporal muscle activation of *M. gastrocnemius medialis* and *soleus* (peak activity), the principal component scores were only altered in the patients with ankle osteoarthritis and the short-term follow-up patients. However, both studies (Chapter 4 and 5) showed that despite remaining changes in the gait patterns, patients had less pain, higher functional ankle scores, and a better general health after supramalleolar osteotomies. Thus, joint preserving realignment surgeries are a promising alternative treatment for asymmetric ankle osteoarthritis.

Chapter 1

Introduction

1. Ankle osteoarthritis

Osteoarthritis (OA) is a degenerative joint disease that is characterized by a degeneration of the articular cartilage, sclerosis of the subchondral bone, the formation of marginal osteophytes, and subchondral cysts [1-3]. The pathological pathways leading to OA are diverse and not entirely understood. One common factor is an increased intra-articular mechanical stress that exceeds the physiological limit (e.g. due to obesity or mal-alignment of the joint). The failure to repair damages from such excessive mechanical stress finally leads to OA. The changes that are observed in OA are an adaptation of the body to the increased stresses that aim to heal the joint and improve the joint biomechanics [1].

It has been estimated that OA affects about 10 % of the world's population aged over 60 [4-7]. The amount of affected people increases with increasing age, since the progress of the disease is not reversible [6]. About 40 % of the OA patients suffer from knee OA, 30% from hand OA, and 20 % from hip OA, while only about 5 % suffer from ankle OA [8]. Other studies showed that less than 10 % of the patients coming to orthopaedic hospitals with lower limb OA suffered from ankle OA [8,9]. An important characteristic of ankle OA is that it is often accompanied by a mal-alignment of the hindfoot. Valderrabano et al. [10] showed that in half of the ankle OA cases a varus mal-alignment of the hindfoot is present. Only 10 % of the patients have a valgus mal-alignment while the other 40 % of the patients have a neutral alignment of the hindfoot.

1.1. Etiology of ankle osteoarthritis

OA can be classified according to the underlying mechanism leading to the disease. If there is no identifiable cause for the OA, one speaks of primary or idiopathic OA. If there is a prior known disease or event that is related to the development of OA, one speaks of secondary OA. Examples for secondary OA are rheumatoid arthritis and posttraumatic OA [2,11]. While knee and hip OA are primarily idiopathic, ankle OA is in more than 70% of the cases posttraumatic [9,10,12]. The most common posttraumatic etiologies were ankle fractures (rotational fractures, malleolar fractures) and ligament ruptures (with either persisting pain or persisting instability) [9,10]. Since these studies were retrospective, the true prevalence of ankle OA after such injuries is unknown. An important consequence of the predominantly posttraumatic nature of ankle OA is that it often affects younger patients.

1.2. Conservative treatment

Conservative or non-surgical treatment options of ankle OA include medication (e.g. pain medication, intra-articular injection of hyaluronic acid), shoe modifications, bracing, use of assistive devices such as canes or crutches, stretching, mobilization, and strengthening [13,14].

Braces or ankle-foot orthoses aim to control and limit the painful motion of the ankle joint in the sagittal plane and to maintain the ankle in a neutral position. Custom made ankle-foot orthoses gauntlet braces provide stability of the hindfoot and ankle joint, while they allow

motion in the forefoot [15]. Other options include the addition of a rocker bottom sole on the patient's shoe in order to facilitate the sagittal plane movement. However, there are no clinical trial data available that describe the effect of braces or orthoses in the treatment of ankle OA[14,15].

The effect of hyaluronic acid injections in ankle OA is discussed controversially. There is evidence that intra-articular injections of hyaluronic acid reduce pain and improve clinical scores [16,17]. However, two randomized controlled trials showed improvements in both the active group (hyaluronic acid) and the placebo group (saline) [18,19]. The follow-up time in all these studies was with maximally 6 months relatively short, and results on the long-term effect are lacking [14].

1.3. Surgical treatment of ankle osteoarthritis

Currently there are two common surgical treatment options for ankle OA: ankle arthrodesis (fusion) and total ankle replacement (TAR) [20,21]. Ankle arthrodesis has been performed for a long time and has historically become a gold standard in the treatment of ankle OA. It provides a good functional outcome with a low complication rate [20,21]. Although approximately 75% of the patients reported a good outcome [22], some drawbacks of ankle arthrodesis due to the restricted mobility of the ankle exist. The risk of OA in the adjacent joints is increased, the gait pattern is changed and the functionality of the joint is limited [21,23]. In contrast to ankle arthrodesis, TAR aims to preserve the function and mobility of the ankle joint, therefore improving the gait pattern [24]. While the first ankle prostheses often failed, recent studies showed a survival rate of around 90% 5 years after surgery [25] and of around 80% 10 years following surgery [22,26]. This limited longevity of TAR is one major concern, however, up to 80% of the patients reported a good outcome [22].

In recent years, realignment surgeries have been proposed as an alternative treatment in earlier stages of the disease where at least 50 % of the joint cartilage is preserved [27-35]. Ankle OA often has an asymmetric nature with a mal-alignment of the hindfoot and a deviation of the joint loading axis into varus or valgus. Clinically, varus mal-alignments of the hindfoot are more frequent [10]. Realignment of the ankle joint is achieved by supramalleolar osteotomies and hindfoot osteotomies. The aim of the surgery is to improve the congruency of the joint, to unload the degenerated cartilage and thus to restore the joint biomechanics [28,35].

1.4. Clinical changes after realignment surgery

The clinical outcome of the different treatments of ankle OA is mainly analyzed by evaluating different scores. The most commonly used clinical score is the American Orthopaedic Foot and Ankle Society (AOFAS) ankle hindfoot score [36,37]. The AOFAS score assesses the level of pain, the function, and the alignment of the joint [38]. Other often used scores are the visual analogue scale (VAS) to assess the level of pain, and the short-form-(SF)-36 questionnaire. The SF-36 measures the health of the patients and is divided into eight dimensions within different areas: functional status (physical functioning, social functioning, role limitations

physical, and role limitations emotional), wellbeing (mental health, vitality, and pain), and general health perception [39,40].

The clinical outcome of realignment surgeries in patients with varus or valgus ankle OA showed a significantly lower level of pain, and significantly higher AOFAS ankle scores [28,29,33]. The ankle scale of Takakura et al. [32] also showed an overall improvement, as well as in the categories pain, walking, and activities of daily living. The range of motion (ROM) of the ankle joint on the other hand remained the same or even slightly decreased [30-32]. Another study however, showed an improvement in the postoperative ankle dorsi- and plantarflexion ROM [41]. It was also seen that postoperatively, more patients participated in recreational sports [41].

2. Gait analysis

Human walking already drew the research interest of early scientist such as Aristotle (384–322 BC). However, only the development of modern computers made studying and analyzing the human gait widely possible [42]. The following four disciplines of science - kinematics, kinetics, electromyography (EMG) and engineering mathematics - are strongly associated with the instrumented gait analysis [43]. Kinematics describe the motion of a body without considering of the cause of the movement. In gait analysis, the human body is modeled with a set of rigid segments. The position of each of these segments is defined by at least three points (markers). Joint angles are then derived by calculating the relative rotation of one segment with respect to the reference segment [44]. Kinetics describe the forces that act on a body and that cause a movement. To study the gait kinetics, force plates are used to measure the ground reaction forces (GRF). Inverse dynamics then allow calculating joint moments and powers from the position of the body segments and the GRF [43]. EMG is used to study the muscle activation during walking. The stimulation of muscle fibers from arriving action potentials result in a change of the membrane potential from a resting potential of around -90 mV to a peak depolarization potential of around 40 mV [45-47]. This change in the membrane potential can be detected with surface electrodes and therefore provide a measure of the muscle activation during the studied task. Finally, engineering mathematics are essential for the data analysis whether that is inverse dynamics to calculate the joint kinetics or a principal component analysis (PCA) for further data analysis.

2.1. Foot models

The conventional lower body model used in clinical gait analyses only models the foot as a single segment [44]. However, this doesn't reflect the anatomy of the foot with its several bones and joints very well. With the emergence of better and more accurate cameras, several new foot models were developed [48-52]. Most of these models capture the tibia, hindfoot, forefoot and hallux, while the more complex ones also track the midfoot or even the medial and lateral forefoot [53]. The Oxford foot model, which was used in combination with the conventional gait model for this thesis captures the tibia, hindfoot, forefoot, and hallux with 34 reflective

markers that are placed on specified anatomical landmarks (Table 1, Figure 1) [48]. The output from the model yields the 3-dimensional hindfoot-to-tibia angles (plantar-/dorsiflexion, inversion/eversion, internal/external rotation), forefoot-to-tibia angles (plantar-/dorsiflexion, supination/pronation, abduction/adduction), forefoot-to-hindfoot angles (plantar-/dorsiflexion, supination/pronation, abduction/adduction), and hallux-to-forefoot angles (plantar-/dorsiflexion).

Table 1: Name and position of the markers used for the Oxford foot model [48] in combination with the conventional gait model [44].

Marker name	Anatomical position	Segment
SACR [a]	Sacrum	Pelvis
LASI, RASI [a]	Anterior superior iliac spine	Pelvis
LTHI, RTHI [a]	Wand marker on mid-thigh (laterally)	Femur
LKNE, RKNE [a, b]	Femoral condyle	Femur, Tibia
LTUB, RTUB [b]	Tibial tuberosity	Tibia
LHFB, RHFB [a]	Head of fibula	Tibia
LTIB, RTIB [a]	Wand marker on mid-shank (laterally)	Tibia
LSHN, RSHN [b]	Anterior aspect of shin	Tibia
LANK, RANK [a, b]	Lateral malleolus	Tibia
LMMA, RMMA [b]	*Medial malleolus*	*Tibia*
LHEE, RHEE [a, b]	Posterior distal aspect of heel	Hindfoot
LCPG, RCPG [b]	Wand marker on posterior calcaneus	Hindfoot
LPCA, RPCA [b]	*Posterior proximal aspect of heel*	*Hindfoot*
LLCA, RLCA [b]	Lateral calcaneus	Hindfoot
LSTL, RSTL [b]	Sustentaculum tali	Hindfoot
LP1M, RP1M [b]	Base of first metatarsal	Forefoot
LD1M, RD1M [b]	Head of first metatarsal	Forefoot
LP5M, RP5M [b]	Base of fifth metatarsal	Forefoot
LD5M, RD5M [b]	Head of fifth metatarsal	Forefoot
LTOE, RTOE [a, b]	Between heads of second and third metatarsals	Forefoot
LHLX, RHLX [b]	Base of hallux	Hallux

Names in italics indicate markers that are only used for the static trial. They are removed during the dynamic capturing.
[a]: used for the conventional gait model
[b]: used for the Oxford foot model

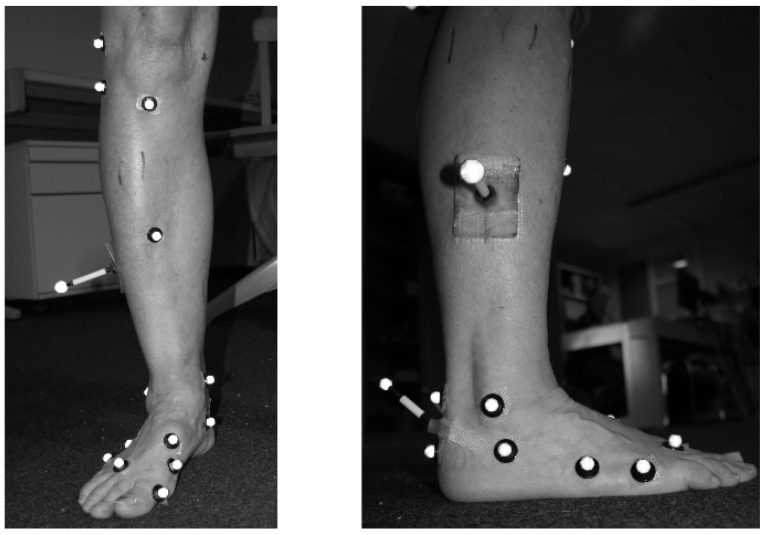

Figure 1: Marker placement on the lower leg according to the Oxford foot model

2.2. Electromyography

Surface EMG provides a non-invasive tool to study the muscle activation during different tasks. In order to achieve reproducible results, standardized procedures are desirable for the recording of surface EMGs. The "Surface EMG for a Non-Invasive Assessment of Muscles" (SENIAM) project provides such standardized recommendations for the preparation of the measurement as well as for the positioning of the surface electrodes [54]. These recommendations are based on the principle that the most reproducible signal is achieved when the electrode is placed "far away" from both the innervation zone and the myotendonous junction. The electrode placement on the lower leg muscles that was used for this thesis is summarized in Table 2.

Table 2: Electrode positioning on the lower leg muscles according to the guidelines of the SENIAM project [54].

Muscle	Electrode position and orientation
M. gastrocnemius medialis	Most prominent bulge of the muscle in direction of the long axis of the leg.
M. gastrocnemius lateralis	1/3 on the line between the head of the fibula and the heel.
M. soleus	2/3 on the line between the medial condyle of the femur to the medial malleolus.
M. peroneus longus	1/4 on the line between the head of the fibula to the lateral malleolus
M. tibialis anterior	1/3 on the line between the tip of the fibula and the medial malleolus.

Besides the location of the innervation zone, there are several other factors that influence the signal and need to be considered in the interpretation [55]. The choice of the size and shape of the electrodes influences the number of active motor units that can be detected. The location of the electrode with respect to the muscle belly influences the amount of possible cross-talk from adjacent muscles. Physiological factors that influence the EMG signal are the number of active motor units, the muscle fiber types, the fiber diameter, the blood flow in the muscle, the depth and location of the active muscles fibers, and the amount of tissue (e.g. fat tissue) that lies between electrode and active muscle fiber [55].

Two characteristics of the EMG signal that are mainly analyzed are the amplitude and the spectral properties. The amplitude increases with increasing force generation [55,56]. However, there are also other factors that influence the amplitude and one cannot estimate the force production solely from the amplitude. Since there are several factors that influence the EMG signal the amplitude is usually normalized to compare EMG amplitudes between different subjects and measurements [55,57,58]. Commonly used reference values for the normalization are the amplitude of the maximal voluntary isometric contraction [58], and the peak or mean amplitude of a dynamic contraction [47,57,58]. The spectral properties of an EMG signal include power spectrum, mean, and median frequency. Similar to the amplitude, the mean frequency increases with increasing torque production [56]. The mean or median frequency is also used to study fatigue since it decreases when the muscle fatigues [47,55]. Conversely discussed is whether the spectral properties of the EMG signal are related to the recruitment of type I and II muscle fibers [59,60]. In patients with OA it has been speculated that the observed lower EMG frequencies are related to an atrophy of mainly type II fibers [61-63].

2.3. Functional changes in the gait patterns of patients with ankle osteoarthritis

Pain and the reduced mobility of the ankle joint complex also led to different changes in the gait patterns of the patients. Compared to healthy controls, patients with end-stage ankle OA walked slower, with a lower cadence, and with shorter strides [64,65]. Furthermore, the range of motion (ROM) of the ankle of the affected side was reduced compared to the non-affected side [66] and compared to healthy controls [65]. One study with a more detailed foot model showed that the ROM of all foot segments (hindfoot, forefoot and hallux) was reduced over a gait cycle as well as within different phases of the gait cycle [64]. Besides the limitation in the mobility, ankle OA patients also showed reduced values in the ankle kinetics. With their affected side, they produced a lower external ankle dorsiflexion moment, absorbed and generated less ankle power [65,66].

Patients with ankle OA also have lower calf circumferences of their affected leg than of the non-affected leg [61]. This is a sign of an atrophy of the calf muscles. This atrophy led to several changes in the affected muscles. End-stage ankle OA patients produced lower maximal isometric ankle plantar- and dorsiflexion torques than healthy controls. Furthermore, the mean frequency of the measured EMG signals of *Mm. tibialis anterior, gastrocnemius medialis, and soleus* was significantly lower in the affected leg than in healthy controls [61]. During walking, this resulted in muscle activation patterns that spanned a broader time range and that contained more low frequency components [67].

2.4. Influence of treatment on joint biomechanics

The effects of ankle arthrodesis on the gait pattern were already studied in the late seventies [68], later several other studies followed. Compared to healthy controls, patients with an ankle arthrodesis walked slower, with a lower cadence, and with shorter strides [68,69]. While all studies showed a reduced sagittal ROM of the ankle or hindfoot compared to controls [69-72], the effect on other joints of the lower extremity was more controversial. Thomas et al. [69] found a decrease in the flexion/extension ROM of the hip, while it was equal to the controls' ROM in the study of Mazur et al. [68]. Similarly for the knee flexion/extension ROM, Beischer et al. [72] found an increase, while Mazur et al. [68] found no differences. Within the foot, there is evidence that an increased ROM of the forefoot compensates for the lack of mobility of the hindfoot [70]. A newer study, however, couldn't confirm these results and found also decreased ROMs of the forefoot [69]. None of these results on the ROM during walking came from prospective studies and the values were compared against healthy controls. The preoperative mobility of the forefoot is therefore not known. A radiographic, prospective study, however, showed that the ROM of the subtalar joint and of the combined midfoot increased. This increase was also positively correlated to the score of the SF-36 questionnaire [73]. For the kinetics, it was found that after arthrodesis patients generate a lower ankle power [72] and had lower maximal anterior, posterior, and vertical ground reaction forces (GRF) [70].

Following a minimum of one year after TAR, patients had an improved walking speed, cadence and stride length. Their dynamic ROM during walking was also increased for the

ankle, knee, and hip joint in the sagittal plane [65,74]. Compared to healthy controls, these values of the ankle ROM were still reduced [65] or equal [75].

Together with persisting muscle atrophy, ankle arthrodesis resulted in a changed temporal activation of *M. soleus* during walking. Contrary to healthy controls, it was not active between loading response and terminal stance but during pre-swing [70]. One year after TAR, patients produced higher isometric plantar- and dorsiflexion torques. However, the lower maximal calf circumference and lower mean frequency of the EMG signals were still seen in the patients, thus indicating that the muscle atrophy didn't recover [62]. Another patient group that was measured on average 3.5 years after TAR (range: 11 – 126 months) showed differences in the temporal muscle activation during walking. *M. gastrocnemius medialis* had a higher activity in early stance and *M. tibialis anterior* a higher activity in terminal stance compared to healthy controls [75].

3. Computational methods

3.1. Wavelet transformation

EMG signals contain information on both time and frequency. One would often like to analyze both these EMG characteristics together. However, due to the uncertainty principle it is not possible to have both infinite time and frequency resolution [76]. Common EMG analyses therefore mainly consist of the analysis of either amplitude characteristics (e.g. by rectifying and smoothing the signal), or by the analysis of the frequency content (e.g. Fourier transformation) [58]. The Fourier transformation requires stationary signals and is therefore not suited for the analysis of dynamic signals such as EMGs that are derived during walking. For such dynamic EMG signals von Tscharner proposed a wavelet transformation with a filter bank of non-linearly scaled wavelets [77,78]. Each of these wavelets is characterized by its center frequency, time resolution, and bandwidth (Table 3) and serves as a band-pass filter for the EMG signal. The wavelets were scaled so that the power of the EMG is retained and the time resolution of the wavelets was adjusted to the physiological properties of the muscle [77,78]. The wavelet transformation yields an intensity pattern that contains time, frequency, and intensity information. It can be depicted in a contour plot where the abscissa represents the time axis, the ordinate the frequency axis, and the grey shading the intensity (Figure 2b). The summation over the time axis yields the total intensity (Figure 2c), while the summation over the frequency axis yields the power spectrum (Figure 2d).

Table 3: Characteristics of the wavelets that were used in this thesis. They were calculated according to von Tscharner [77].

	w_0	w_1	w_2	w_3	w_4	w_5	w_6	w_7	w_8	w_9	w_{10}	w_{11}	w_{12}
Center frequency (Hz)	7	19	38	62	92	128	170	218	271	331	395	466	542
Bandwidth (Hz)	12	22	30	39	47	59	66	76	84	94	101	111	118
Time resolution (ms)	80	53	39	30	25	21	19	17	15	13	13	11	11

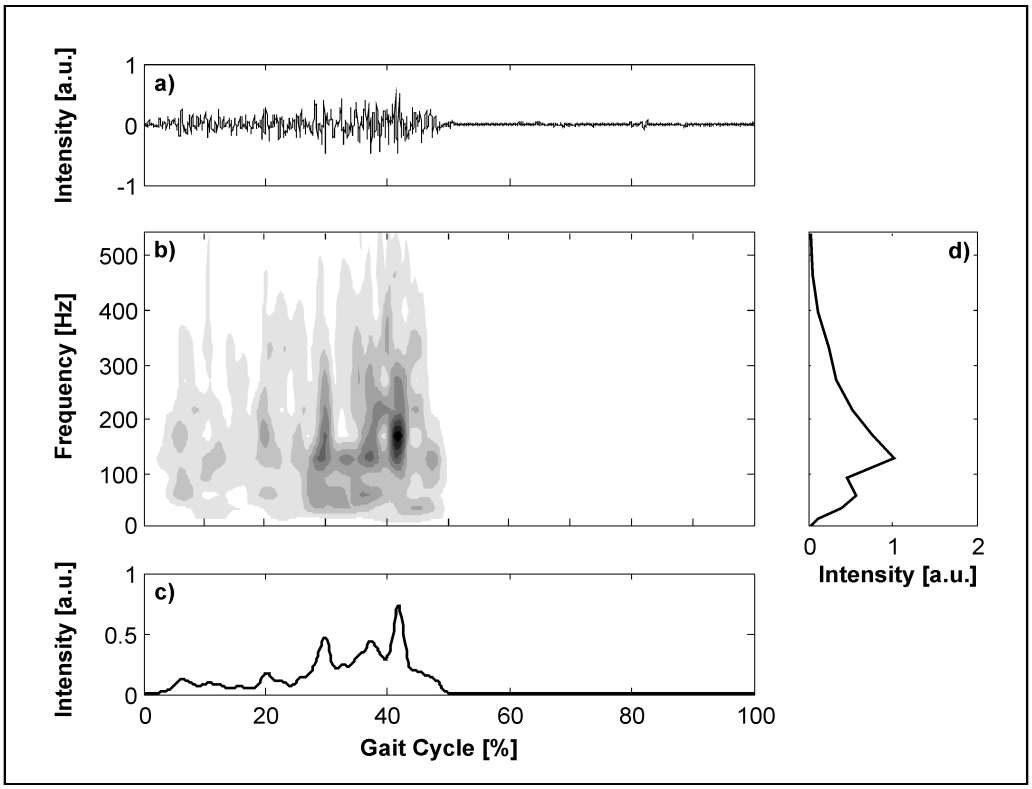

Figure 2: Example of the processing of a raw EMG signal (a) from *M. gastrocnemius medialis* during one gait cycle. The grey scale of the wavelet pattern (b) indicates the intensity, with darker colors indicating higher intensities. The total intensity (c) and the power spectrum (d) are derived by summing over the time axis, or frequency axis, respectively.

The wavelet transformation has been successfully applied in the analysis of EMG signals. It was used to investigate gender differences between the muscle activation of runners [79], for the classification between ankle OA patients and healthy controls [67], as well as between fatigued and non-fatigued runners [80]. Other studies used a continuous wavelet transformation to study spectral properties such as the instantaneous mean frequency of EMG signals during dynamic contractions (e.g. [56,81,82]).

3.2. Principal component analysis

Principal component analysis (PCA) is a statistical method that can be used for data reduction. It is an orthogonal transformation that maximizes the variance that is explained by each component (Figure 3). It is characterized by a set of new, transformed axes (principal component (PC) vectors), and the loadings of each data point on these axes (PC scores). The PC vectors are orthogonal to each other and therefore uncorrelated. They are sorted according to the variance they explain and since the variance that each component explains is maximized, it is possible to summarize the original input data with only a few components [83,84].

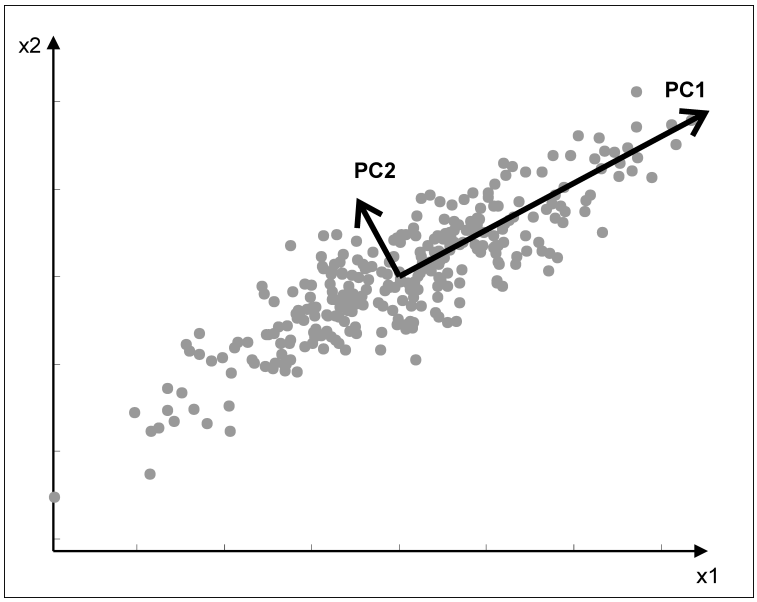

Figure 3: Example of the two principal components (PC) of a two-dimensional data set (x1, x2).

One application of PCA in the analysis of gait data is the quantification of differences in the waveform data of different subjects. Waveform data (e.g. joint angles, or moments) are organized in a matrix X (m data points x n subjects or trials). This matrix X is then transformed into a matrix Z that contains the uncorrelated PC scores. Additionally, the eigenvectors U of the covariance matrix of X form the new axes of the transformed data, the PC vectors, and the eigenvalues λ indicate the amount of variance that is explained by one component. The original data can be reconstructed from the principal components by solving $X = UZ$. Similarly, it is also possible to reconstruct the data by only using a single PC, thus showing the effect of this particular PC vector and PC score. In order to achieve a data reduction, only the first few PCs that explain most of the variance are retained. There are several possibilities of determining the number of retained PCs. One is to retain a proportion of the total variance, e.g. 90 or 95% [83-85]. Another possibility is to use the broken stick rule $b_k = \frac{1}{p}\sum_{i=k}^{p}\frac{1}{i}$: If the eigenvalue of the kth component is higher than b_k then the component should be retained [86,87].

PCA became increasingly popular in the analysis of time series data of different movements (joint angles or moments), so called waveform data. An advantage of PCA is that it doesn't involve any a-priori parameter selection. Gait waveform data are usually normalized to a gait cycle and then submitted to the PCA. The PC vectors represent features of the original data and indicate where the subjects' waveforms differ. The PC scores are the individual loadings on each PC vector that are needed to reconstruct the original waveforms. For each principal component and input trial, one PC score is obtained that can be used for statistical testing of group differences. Both PC vectors and scores can therefore be used to characterize the gait patterns of different patient groups, as well as the effect of treatments. In knee OA it was shown that patients have both a lower magnitude and range of motion in knee flexion, a lower amplitude and magnitude of the knee flexion moment and a higher magnitude and amplitude of the knee adduction moment [85]. Additionally, it was seen that changes in the gait patterns

depend on the severity of knee OA [88,89] and the gender of the patients [90]. Total knee replacement resulted in an improved gait pattern, i.e. closer to a healthy gait pattern, for kinematics, kinetics [91], and temporal muscle activation [92]. Besides gait patterns of OA patients, PCA was also used for the description of gait changes in other patient groups. For children with autism it was seen that they walk with a decreased ankle plantarflexion moment during the first half of the gait cycle and with an increased ankle dorsiflexion angle throughout the whole gait cycle [93]. Patients that had a lower limb fracture showed changes in the first PC of the vertical GRF compared to healthy controls. This resulted in a lower GRF at the end of the step with the affected leg and in a higher GRF at the beginning of the step with the healthy leg. After additional physiotherapy treatment most of the treated patients moved within the boundaries of the vertical GRF pattern of the healthy controls [94]. PC scores from the 3-dimensional GRF were also used to assess influences of medication and/or stimulation on the gait pattern of patients with Parkinson. Without medication and stimulation, none of the patients was within the boundaries of a normal pattern, while the best result was found with both medication and stimulation with four of 10 patients within the boundaries [95].

3.3. Support vector machine

Support vector machines (SVM) are classification tool in pattern recognition that has become increasingly popular recently. Common applications include digit recognition, or face detection [96]. However, SVMs have also found their way to movement analysis, specifically to the classification of gait patterns between different subjects [36,97-102].

The main idea of the SVM classifier is to find a hyperplane that maximizes the margin between the two groups (Figure 4). The data points that form the margin are called the support vectors and a removal of these points would change the solution of the classification. An advantage of the SVM classifier is therefore that data points, which lie far away from the hyperplane don't influence the result of the classification [96,103]. A linearly separable data set is the easiest case. However, the SVM classifier can also be used for linearly non-separable data sets. For the linear SVM a parameter *C* is chosen that determines how many mistakes are tolerated. A larger *C* means that errors have a higher weight and therefore leads to fewer training errors, but also a lower generalization performance. *C* is generally adjusted so that a maximal cross-validation rate for the training data set is achieved [96,103,104]. Another option is the use of nonlinear SVMs. In this case the data points are mapped into a higher dimensional space according to the chosen kernel and the maximization of the margin is performed in this higher dimensional space. The following three kernels are commonly used for SVM classifications [96,97,102,103]:

- Linear kernels: $K(x_i, y_j) = x_i \cdot y_j$
- Polynomial kernels: $K(x_i, y_j) = (x_i \cdot y_j) + 1)^d$ (*d*: degree of polynomial)
- Radial basis functions: $K(x_i, y_j) = \exp\left(-\frac{\|x_i - y_j\|^2}{2\sigma^2}\right)$ (σ: width of Gaussian kernel)

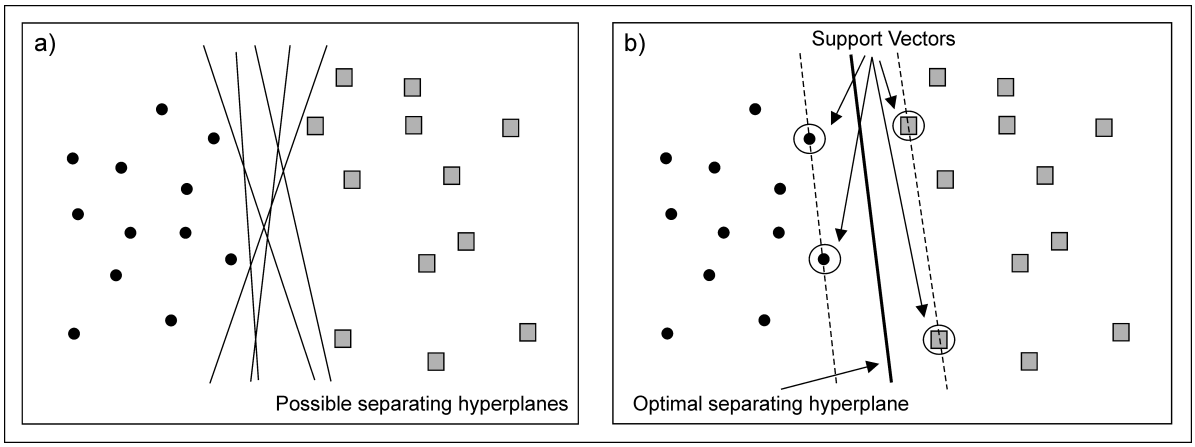

Figure 4: An example showing the separation of two classes (squares and dots) by a) several possible hyperplanes, and b) the optimal hyperplane that maximizes the margin between the two groups. The data points that form the maximal margin (circled) are called support vectors.

These three kernels were also used in the classification of gait patterns. Several studies investigated the different gait patterns of young and elderly people. The classification was either based on specific kinematic and kinetic variables (peak values) [97-99], or on features of the GRF that were extracted by PCA [102]. In the diagnostics of pathological gait patterns cadence and step length have been used to identify the gait of patients with cerebral palsy [105], and features of the GRF and the hindfoot kinematics for the identification of patients with patellofemoral pain syndrome [100]. Another study used SVMs to investigate the recovery after total knee replacements. The spatiotemporal gait variables of healthy subjects and knee OA patients were used as training set for the SVM. Patients were classified again after two and 12 months rehabilitation. While after two months only a few patients were classified as healthy, after 12 months all except two patients were classified as healthy. The results of the classification were also in agreement with the clinical knee score in a way that patients with higher scores were more likely to be classified as healthy [101].

4. Aims and hypotheses of this thesis

Although that the gait patterns of patients suffering from ankle OA have been described before [64-67], the patients that participated in these studies suffered from end-stage ankle OA. Patients with moderate and asymmetric ankle OA might benefit from joint preserving realignment surgery, but it is not known whether they already show the same changes as patients with further progressed disease. Furthermore, the gait patterns of ankle OA patients have so far only been described with ranges of motion in the ankle joint or of different foot segments, or by reporting peak values of the GRF, joint moments, or powers. Recent studies on OA showed that PCA provides a powerful tool that allows the analysis of the measured temporal waveforms of joint angles, moments, powers, or the temporal muscle activation [85,88-92,106]. Therefore one can analyze differences in the waveforms without preselecting the features of interest. Another important factor that was often neglected is that movements are not possible without muscles. However, changes in the muscle activation of ankle OA patients were only rarely reported. Here, the wavelet transformation with its

resulting wavelet pattern enables the analysis of both time and frequency content of the EMG signal during dynamic contractions. Thus, the first aim of this thesis was to characterize the gait pattern of patients with moderate ankle OA and a varus or valgus mal-alignment of the hindfoot, taking into account the temporal waveforms of the kinematics and kinetics, as well as the muscle activation. This involved testing the following hypotheses: (i) the patients have a reduced range of motion of the hindfoot segment and peak kinetic values that result in significantly different PC scores; (ii) the features that can be extracted with the PCA allow a successful classification between ankle OA patients and healthy controls; (iii) patients with moderate asymmetric ankle OA show a shift towards lower frequencies in the lower leg muscle activation that lead to different characteristics of the wavelet patterns than in the controls.

Joint preserving realignment surgery provides pain relief and leads to an improved function as indicated by the improved clinical scores [28-32,41]. However, it is not clear whether the biomechanical and neuromuscular gait pattern improve in a similar way. Further, it is important to know whether the surgery also leads to a long-term beneficial effect or whether the gait patterns worsen over time. Therefore, the second aim of this thesis was to quantify the rehabilitation potential of the biomechanical and neuromuscular gait pattern after realignment surgery, both in a prospective, short-term study and in a cross-sectional, long-term study. In these studies the hypotheses of (iv) a long-term improvement of the spatiotemporal, kinematic, and kinetic gait parameters, and (v) that the features that are resolved by PCA from the foot kinematics and the temporal muscle activation are closer to the ones from healthy controls in the postoperative patients.

5. Outline of this thesis

The results of this thesis are presented in four chapters that cover the following research questions and topics:

Chapter 2: This chapter focuses on the biomechanical changes in the gait pattern of asymmetric ankle OA patients compared to healthy controls. The use of the Oxford foot model in the gait analysis allows a more detailed analysis of the osteoarthritis related changes in the foot kinematics compared to some of the earlier studies [65]. The data analysis combines the conventional approach of reporting peak values, and ranges of motion with a PCA approach that allows the comparison of the actual waveforms. In the end, the results of the gait waveform analysis are used to investigate whether the patients' gait pattern are sufficiently different from healthy controls that a successful classification with a linear SVM is possible.

Chapter 3: The focus of this chapter is on the neuromuscular adaptations to moderate asymmetric ankle OA both during maximal isometric contractions and during walking in comparison to healthy controls. It addresses changes in the frequency content, i.e. wavelet power spectrum, and in the time domain by using a wavelet transformation and introduces the

entropy for the analysis of the wavelet patterns. Further it addresses differences in the muscle activation between patients with a varus or valgus mal-alignment of the ankle.

Chapter 4: This cross-sectional study was done to investigate the long-term effects of supramalleolar osteotomies on the patients' gait pattern. Since this is a clinically orientated paper, the analysis of the gait data was performed in a conventional way of reporting peak values and ranges of motion. Another important aspect of this chapter is that it also addresses secondary changes in the knee and hip joint.

Chapter 5: The last study of this thesis investigates whether the previously observed clinical improvements after supramalleolar osteotomies are also reflected in an improvement of the gait pattern. Contrary to chapter 4, the gait patterns were analyzed by PCA. The results are divided into two parts. Part one focuses on the description of the postoperative gait pattern in both short-term follow-up and long-term follow-up patients. The second part only addresses changes in those patients that were measured prospectively.

Chapter 6: The last chapter contains a general conclusion of the results of this thesis. It discusses both methodological aspects and the clinical relevance of this research project. Further, it gives an outlook on possible future projects.

Since this thesis is based on individual and independent journal articles, there are some repetitions in the chapters, especially regarding the introduction and the methods.

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Chapter 2

Gait patterns of asymmetric ankle osteoarthritis patients

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An adapted version was published in: Clinical Biomechanics. 2012; 27(6):613-618

Background: In early stages, ankle osteoarthritis is often asymmetric with only partially degenerated joint surfaces. There is only limited knowledge on the effect of asymmetric ankle osteoarthritis on the patients' gait patterns. Therefore, the aim of this study was to characterize kinematic and kinetic changes compared to healthy adults.
Methods: Instrumented gait analysis was performed in eight asymmetric ankle osteoarthritis patients and 15 healthy controls. Beside conventional gait analysis methods, principal component analysis was used to analyze temporal progression of the most important variables: hindfoot dorsiflexion angle and vertical ground reaction force.
Findings: Asymmetric ankle osteoarthritis patients had a lower hindfoot dorsiflexion and rotation range of motion as well as reduced peak ground reaction forces and peak kinetic values. Principal component analysis revealed that for both the hindfoot dorsiflexion angle and the vertical ground reaction force those principal component vectors affecting the amplitudes had significantly lower principal component scores in patients than in controls. The use of the principal component scores for classification with a linear support vector machine resulted in a high recognition rate of 97.8% for the discrimination between the affected leg and the healthy controls.
Interpretation: Patients with asymmetric ankle osteoarthritis suffer from substantial pathological kinematic and kinetic gait changes. Principal component analysis combined with a linear support vector machine could successfully be used to temporally quantify and classify asymmetric ankle osteoarthritis gait patterns. This study therefore helps to understand the pathomechanism of early stage ankle osteoarthritis from a biomechanical view.

Keywords: ankle osteoarthritis; gait analysis; hindfoot; ground reaction force; principal component analysis

1. Introduction

Osteoarthritis (OA) in the ankle joint typically develops after trauma (e.g. malleolar fracture or ankle ligament lesions) and often affects younger patients. Although ankle injuries are very common, ankle OA is less prevalent than hip or knee OA [1-3]. At an early stage, ankle OA is often asymmetric and only a part of the joint surface is involved [3]. Over time the asymmetric ankle OA develops to a full end-stage ankle OA. Patients with end-stage ankle OA suffer from pain and reduced mobility of the ankle joint complex. During level walking such a pain-caused arthrogenous muscle inhibition leads to altered muscle activity patterns [4]. Gait patterns of end-stage ankle OA patients are characterized by an alteration of muscle activities, a decrease of the range of motion (ROM) of the ankle joint complex throughout a gait cycle, and lower peak values of the ground reaction forces (GRF), ankle moments, and ankle power [4-7].

In contrast to end-stage ankle OA, in-vivo biomechanical data on early asymmetric ankle OA are, to our knowledge, missing in the literature. Biomechanical studies in early asymmetric ankle OA are therefore necessary to characterize the pathomechanisms in this stage of the disease where surgical joint preserving treatment options to slow down the OA process are more promising. Joint preserving surgery of ankle OA includes corrective osteotomies for joint load redistribution, ligament repair, cartilage repair, and tendon transfers, aiming to avoid surgical treatment options of end-stage OA such as total ankle replacement or ankle joint fusion [8].

Conventional analysis of biomechanical gait data often includes reporting data by ranges, minimal, and/or maximal values, as well as time when these events occur. However, this kind of analysis neglects the temporal progression of joint angles and kinetics during gait, which is subsequently lost by the data reduction. A newer method in the analysis of gait patterns is the principal component analysis (PCA), which has the advantage of retaining temporal information. PCA was used to characterize changes in the gait patterns of knee OA patients [9-11], or to study the GRF of patients with lower limb fractures [12]. The application of a PCA on gait data allowed successful classification of different groups using for example linear discriminant functions [10,11] or support vector machines (SVM) [13]. SVMs are supervised learning algorithms that are used to classify data sets by maximizing the margin between two groups [14]. In case of OA patients, linear SVM could classify the spatiotemporal parameters of knee OA patients and healthy subjects with a high recognition rate of 100% in training and 88.9% in testing [15].

The aim of this study was to provide missing biomechanical gait data for unilateral asymmetric ankle OA patients in comparison with healthy subjects. We hypothesized, that (i) patients suffering from asymmetric ankle OA show gait patterns with reduced hindfoot ROMs, peak GRFs, and peak kinetic values; (ii) principal component (PC) scores of the hindfoot dorsiflexion angle and the vertical GRF are significantly different in the patients; and (iii) features from the PCA are sufficiently different that an SVM classifier can discriminate between patients and controls.

2. Methods

2.1. Subjects

The patient group consisted of eight patients (4 male, 4 female; mean age: 53.4 (SD: 11.4) years; mean body mass: 86.1 (SD: 16.3) kg; mean height: 1.75 (SD: 0.08) m) suffering from posttraumatic unilateral asymmetric ankle OA, i.e. with only partially osteoarthritic ankle joint surface (less then 50% of the joint; mean OA grade 1.5, range 1-2, as graded by Morrey and Wiedeman [16]: grade 1: mild narrowing of joint space with mild osteophytes; 2: moderate narrowing and osteophytes; 3: advanced joint degeneration with loss of ankle joint space). The patients were recruited from our orthopaedic outpatient clinic and were assessed prior to their planned joint preserving surgery, i.e. realignment surgery with corrective distal lower leg osteotomies, ankle ligament, cartilage, and tendon repairs [8]. Diagnosis of asymmetric ankle OA was made by x-ray and confirmed intraoperatively. The patients' clinical status on function, alignment, and pain was determined by the American Orthopaedic Foot and Ankle Society (AOFAS) score (mean score: 60.5 (SD: 16.6) points) [17] and their pain level on a visual analogue scale ranging from 0 to 10 (mean: 5.2 (SD: 2.0)). Exclusion criteria were diabetes mellitus, neurological disorders, full end-stage ankle OA, early unilateral ankle OA with symmetric alignment pattern, and asymmetric ankle OA with pathologies or posttraumatic entities of the contralateral leg.

The control group consisted of age-matched 15 healthy subjects (9 male, 6 female; mean age: 48.5 (SD: 10.5) years; mean body mass: 74.6 (SD: 9.9) kg; mean height: 1.73 (SD: 0.12) m) that were pain free and without history of pathologies or surgeries on the lower limbs. Therefore they had a normal clinical-functional AOFAS score (100 points). Controls had a significantly lower body weight than the patients, whereas age and height showed no significant group differences. The local ethics committee approved the study and all subjects gave informed consent before participating.

2.2. Gait analysis and data processing

Gait analysis was performed using a six camera motion capture system (Vicon MX13+, Oxford, United Kingdom, sampling rate: 120 Hz) and two force plates (Kistler, Winterthur, Switzerland, sampling rate: 2400 Hz). Since the Helen Hayes model [18] only captures the foot as one segment, but in the Vicon plug-in also calculates ankle kinetics, it was combined with the Oxford foot model [19] that calculates different foot segments, e.g. the hindfoot. Therefore the use of both models as plug-ins in the Vicon Nexus software resulted in the following analyzed variables: three dimensional (3D) ankle joint moments, and ankle joint power, as well as 3D hindfoot kinematics (motion of hindfoot relative to tibia).

First, a static calibration trial in neutral stance position was recorded in order to reference the hindfoot angles to the neutral position [20]. After the static trial several foot markers were removed as described by Stebbins et al. [19]. Then the subjects completed six walking trials at a self-selected speed during which the subjects cleanly hit the force plates otherwise the trial was repeated. Left and right gait cycles were analyzed separately, leading to 12 trials per subject.

In each subject the gait cycles from the affected or the dominant (defined as the preferred hopping leg [21]) leg were identified. A stance phase symmetry index was calculated by dividing the stance phase percentage of the affected leg (AFL) by the one of the non-affected leg (NAL) and of the non-dominant leg by the dominant leg, respectively.
After determining the ROM of the hindfoot angles and the maximal values of the kinetic variables, all gait data were then normalized to one gait cycle with 101 values (0 to 100%), starting and ending at heel strike. The progressions over time of the normalized data were called waveforms.

2.3. Principal component analysis

As most of the ankle joint complex movement occurs in the sagittal plane, only the hindfoot dorsiflexion angle was analyzed. Similarly, for the GRF only the vertical component was analyzed. For each of the two waveforms the mean over all 276 (23 subjects x 12 trials) was calculated and then subtracted from each trial. These new waveforms were then separately organized in matrices X (276 x 101) and submitted to a PCA. PCA is an orthogonal transformation, which maximizes the variance that is explained by one PC. The eigenvectors of the covariance matrix of X, the so-called PC vectors, represent the orthogonal axes of the new vector space. The PC scores are obtained by projecting an individual trial onto the PC vectors. The eigenvalues represent the amount of the variance that is explained by each PC. The sum of all eigenvalues corresponds to 100% of the variance [10]. In order to achieve a data reduction, of all PC vectors only the first ones explaining 90% of the total explained variance were used [10,22].

2.4. Support vector machine

The retained PC scores of the hindfoot dorsiflexion angle and the vertical GRF (Table 2) were used for classification with a linear SVM [23]. The linear SVM models were trained over the range of C = (0.01 0.1, 1, 10, 100, 1000) and the model with the best recognition rate was selected for classification. The expected recognition rate for data from a new subject was calculated using a leave-one-out cross-validation procedure.

2.5. Statistics

The trials were divided into four groups: dominant leg of the controls, non-dominant leg of the controls, AFL and NAL of the patients. Data were averaged for each subject and group. Due to the small sample size, a Wilcoxon rank sum test with a significance level $\alpha = 0.05$ was used to test for group differences in the subject means between the non-dominant leg and the AFL, and between the dominant leg and the NAL.

3. Results

3.1. Gait characteristics

Table 1 gives an overview on the gait characteristics. Asymmetric ankle OA patients walked with a significantly lower cadence and speed than the controls. They also had a less symmetrical gait pattern, as their stance phases were of significantly different length. The hindfoot dorsiflexion ROM of both the AFL and the NAL of the patients was significantly lower than in the controls. For the kinetics, the positive peak of the anterior-posterior GRF (breaking) and the first active peak of the vertical GRF, as well as the ankle peak moment and peak power were significantly lower in the AFL of the patients than in the controls. Patients also had a lower ankle peak dorsiflexion moment on their NAL. The vertical GRF also showed significant differences in the timing of the second active peak. It occurred significantly later in the gait cycle on the AFL, and earlier in NAL compared to the controls (Table 1).

Table 1: Gait characteristics of the two subject groups

	Leg	Control	Patients	*P* Value
		Mean (SEM)	Mean (SEM)	
Cadence (steps/min)		112.27 (1.52)	102.98 (1.89)*	0.022
Walking speed (m/s)		1.26 (0.02)	1.10 (0.04)*	0.031
Stance phase (%)	AFL	60.95 (0.35)	60.23 (0.54)	0.259
	NAL	60.99 (0.36)	62.79 (0.95)	0.100
Symmetry index stance phase		1.00 (0.004)	0.96 (0.01)*	0.015
Hindfoot dorsiflexion ROM (°)	AFL	22.64 (0.97)	17.96 (1.47)*	0.022
	NAL	22.61 (1.06)	18.17 (2.15)*	0.031
Hindfoot inversion ROM (°)	AFL	15.10 (1.18)	14.71 (0.86)	0.821
	NAL	15.93 (1.10)	14.21 (0.99)	0.349
Hindfoot rotation ROM (°)	AFL	12.67 (0.71)	10.15 (1.62)	0.129
	NAL	14.52 (1.34)	10.89 (0.88)	0.114
Maximum medial-lateral GRF (N/kg)	AFL	0.60 (0.03)	0.63 (0.04)	0.540
	NAL	0.59 (0.03)	0.66 (0.66)	0.317
Positive peak anterior-posterior GRF (breaking) (N/kg)	AFL	1.82 (0.11)	1.39 (0.09)*	0.011
	NAL	1.85 (0.11)	1.59 (0.11)	0.208
Negative peak anterior-posterior GRF (propulsion) (N/kg)	AFL	-2.16 (0.08)	-1.62 (0.22)	0.066
	NAL	-2.09 (0.08)	-1.97 (0.11)	0.232
First active peak vertical GRF (N/kg)	AFL	11.17 (0.24)	10.44 (0.20)*	0.022
	NAL	11.21 (0.20)	11.17 (0.24)	0.821
Second active peak vertical GRF (N/kg)	AFL	11.58 (0.16)	10.46 (0.44)	0.076
	NAL	11.54 (0.15)	11.33 (0.29)	0.923
Time of first active peak of the vertical GRF (% gait cycle)	AFL	13.91 (0.43)	15.72 (1.11)	0.165
	NAL	13.98 (0.48)	14.27 (0.64)	0.675
Time of second active peak of the vertical GRF (% gait cycle)	AFL	46.86 (0.40)	41.31 (2.28)*	0.031
	NAL	46.68 (0.34)	48.93 (0.59)	0.006
Ankle peak dorsiflexion moment (Nm/kg)	AFL	1.74 (0.05)	1.43 (0.14)*	0.042
	NAL	1.74 (0.03)	1.61 (0.05)*	0.042
Ankle peak power (W/kg)	AFL	4.72 (0.18)	3.04 (0.61)*	0.011
	NAL	4.74 (0.17)	4.07 (0.38)	0.100

AFL: affected leg of patients, non-dominant leg of controls; NAL: non-affected leg of patients, dominant leg of controls; GRF: ground reaction force; ROM: range of motion; SEM: standard error of the mean
*: statistically significant differences between patients and controls (Wilcoxon test)

3.2. Principal component analysis

A reduction of PC vectors with the cut-off criterion of 90% of the total explained variance led to four PC vectors retained for the hindfoot dorsiflexion angle and to six PC vectors for the vertical GRF. The effects of the different PC vectors are summarized in Table 2.

Table 2: Principal component scores of the hindfoot dorsiflexion angle and the vertical ground reaction force

PC	Explained Variance (%)	Feature	Leg	Control	Patients	*P* Value
				Mean (SEM)	Mean (SEM)	
Hindfoot Dorsiflexion Angle						
PC1	62.4	Magnitude of hindfoot angle throughout gait cycle	AFL	0.32 (8.74)	6.55 (13.86)	0.498
			NAL	-9.40 (7.24)	10.46 (14.02)	0.208
PC2	15.2	Amount of plantarflexion at push off	AFL	0.19 (3.60)	-8.39 (7.75)	0.146
			NAL	2.49 (3.82)	3.38 (6.35)	0.923
PC3	9.2	Timing of minimum at push off	AFL	-3.57 (2.76)	3.31 (3.73)	0.165
			NAL	0.20 (2.61)	3.01 (6.36)	0.723
PC4	5.9	Amplitude	AFL	5.84 (1.80)	-7.33 (2.31)*	0.001
			NAL	5.32 (1.51)	-13.59 (2.77)*	< 0.001
Vertical Ground Reaction Force (GRF)						
PC1	35.6	Timing of GRF decrease at push off	AFL	-0.40 (0.98)	-3.72 (1.43)	0.114
			NAL	-0.34 (0.85)	5.11 (2.06)*	0.036
PC2	28.4	Amplitude	AFL	1.31 (1.04)	-5.87 (1.63)*	0.001
			NAL	1.22 (0.90)	1.13 (1.14)	0.821
PC3	15.4	Height of second active peak	AFL	0.88 (0.57)	-1.73 (1.64)	0.146
			NAL	1.06 (0.58)	-1.90 (1.67)	0.165
PC4	7.5	Timing of mid stance and height of first active peak	AFL	0.31 (0.61)	-0.66 (0.91)	0.420
			NAL	0.07 (0.47)	-0.05 (1.04)	0.923
PC5	3.1	Height of impact and first active peak	AFL	-0.03 (0.39)	0.23 (0.33)	0.974
			NAL	-0.06 (0.42)	-0.06 (0.41)	1.000
PC6	2.5	Height of midstance valley	AFL	-0.05 (0.31)	-0.09 (0.41)	0.771
			NAL	0.01 (0.23)	0.16 (0.61)	0.583

AFL: affected leg of patients, non-dominant leg of controls; NAL: non-affected leg of patients, dominant leg of controls; PC: principal component; SEM: standard error of the mean
*: statistically significant differences between patients and controls (Wilcoxon test)

3.2.1. Hindfoot dorsiflexion angle

In Figure 1A, the mean hindfoot dorsiflexion angle of the non-dominant leg of the controls and the AFL of the patients are depicted. The visual differences were characterized using PCA. The PC1 vector (Figure 1B) has only positive values and thus affects the magnitude of the angle throughout the gait cycle (Figure 1C). It explained 62.4% of the total variance but didn't show any significant group differences. The only feature that showed significant differences between the patients and the controls was the PC4 scores that affect the amplitude of the hindfoot angle. These PC4 scores were lower in both the AFL and NAL of the patients compared to the controls (Table 2 and Figure 1D).

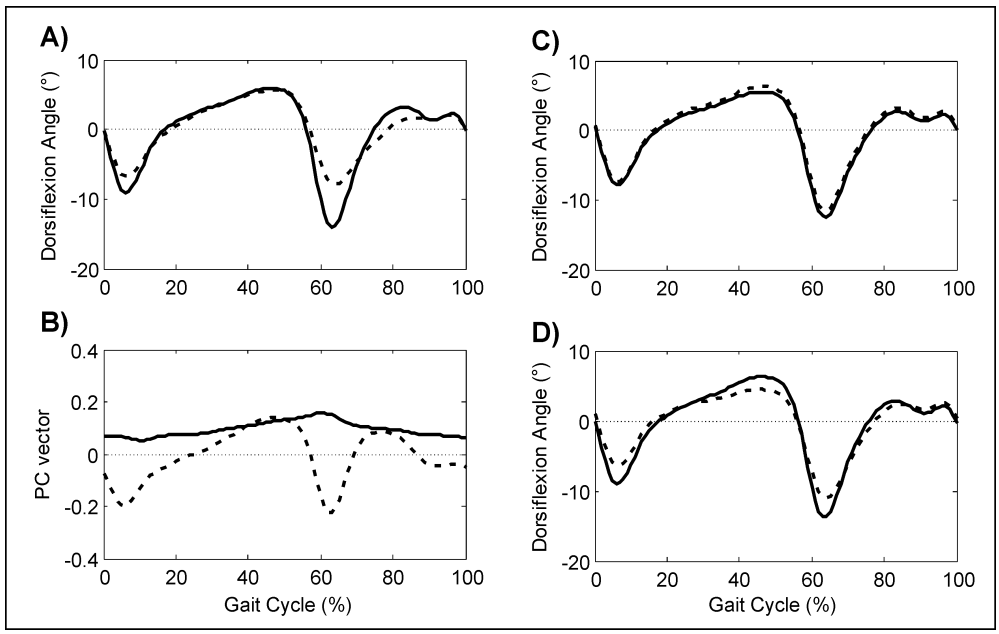

Figure 1: PCA of the hindfoot dorsiflexion angle: A) Mean hindfoot dorsiflexion angle of the non-dominant leg of the controls (solid) and the affected leg of the asymmetric ankle OA patients (dashed). B) First (PC1, solid) and fourth (PC4, dashed) principal component vectors. C) Hindfoot dorsiflexion angles computed by adding the mean PC1 scores multiplied with the PC1 vector to the previously subtracted common mean of the controls (solid) and the OA patients (dashed). D) Hindfoot dorsiflexion angles corresponding to the mean PC4 scores of the controls (solid) and the OA patients (dashed).

3.2.2. Vertical ground reaction force

The temporal behavior of the vertical GRF showed major differences between the AFL of the patients and the non-dominant leg of the controls in the values of the peaks and the valley (Figure 2A). However, the PCA showed that most of the variance (PC1: 35.6%) was caused by the timing of the decrease after the second peak (Figure 2B/C). The significantly higher PC1 scores on the NAL of the patients reflect a lengthening of the stance phase. The PC2 vector (28.5%) changed the amplitude of the vertical GRF waveform and thus affected the height of the peaks and the valley (Figure 2B/D). The corresponding PC2 scores were significantly lower in the AFL of the patients than in the non-dominant leg of the controls (Table 2).

3.3. Classification

Using the retained 10 PC scores (Table 2) from the hindfoot dorsiflexion angle and the vertical GRF in a linear SVM classification resulted in a high recognition rate of 97.8% ($C = 0.1$) between the AFL of the patients and the non-dominant leg of the controls. In order to visualize the classification, the decision hyperplane was projected into the two-dimensional vector space spanned by the hindfoot PC4 scores and the vertical GRF PC2 scores (Figure 3). These two scores were chosen because they showed significant differences between the AFL of the patients and the non-dominant leg of the controls.

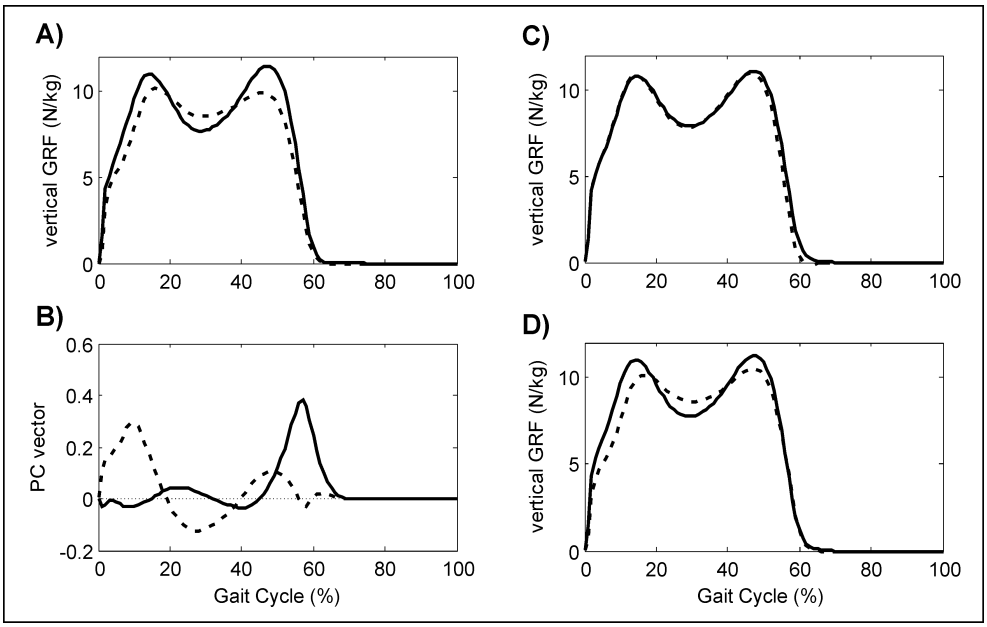

Figure 2: PCA of the vertical ground reaction force (GRF): A) Mean vertical GRF of the non-dominant leg of the controls (solid) and the affected leg of the asymmetric ankle OA patients (dashed). B) First (PC1, solid) and second (PC2, dashed) principal component vectors. C) Vertical GRF computed by adding the mean PC1 scores multiplied with the PC1 vector to the previously subtracted common mean of the controls (solid) and the OA patients (dashed). D) Vertical GRF corresponding to the mean PC2 scores of the controls (solid) and the OA patients (dashed).

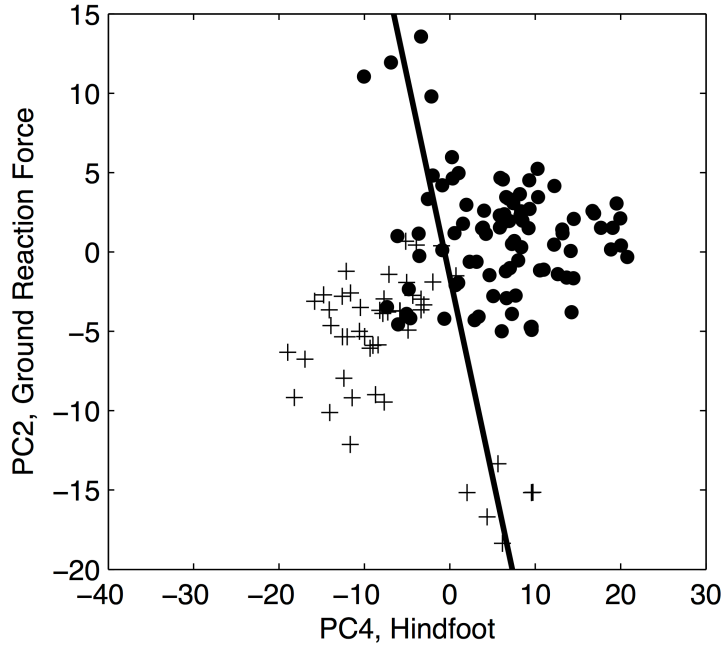

Figure 3: Scatter plot of the hindfoot PC4 scores and the ground reaction force PC2 scores: It shows the scores for each trial of the non-dominant leg of the controls (dots) and the affected leg (crosses) of the ankle OA patients. The decision hyperplane of the linear support vector machine classification was projected into the two-dimensional space of the two selected scores (solid line).

4. Discussion

The results of this study support our hypothesis that patients suffering from asymmetric ankle OA have a decreased hindfoot ROM and peak kinetic values. PCA revealed two main effects that differ between the AFL of the patients and the non-dominant leg of the controls: the PC4 vector of the hindfoot dorsiflexion angle (amplitude throughout the gait cycle) and the PC2 vector of the vertical GRF (amplitude). The use of these two features in a linear SVM classifier resulted in a high recognition rate, thus confirming our third hypothesis.

The gait patterns of the asymmetric ankle OA patients were characterized by a decreased walking speed, cadence, hindfoot dorsiflexion, and rotation ROM. These results are mainly in agreement with previous studies on end-stage ankle OA that reported a decreased walking speed, cadence, and ROM in all three planes of the ankle [5] and hindfoot [6], respectively. As seen in the stance phase symmetry index, asymmetric ankle OA patients had a relatively longer stance phase on the NAL compared to the AFL. This difference in stance phase percentages leads to a limping gait pattern which has been observed in clinical studies in both end-stage [5] and asymmetric ankle OA [8] since the patients spare the painful joint. As in end-stage ankle OA, patients had lower peak values in the 3D GRFs, the ankle dorsiflexion moment, and the ankle joint power of the AFL [5]. In accordance with the differences in the stance phase symmetry index, the timing of the second active peak of the vertical GRF showed significant differences between the two groups. These differences in timing and height of the peak vertical GRFs could further add to a limping gait pattern. For clinical relevance, asymmetric ankle OA seems to have similar influence on gait changes compared to end-stage ankle OA, although large parts of the ankle joint surface are still preserved in asymmetric ankle OA.

PCA proved to be a suitable method to analyze different waveforms throughout the gait cycle. Significant differences were found in the amplitude of the hindfoot dorsiflexion angle which was lower in both the AFL and NAL of the patients. The mean PC4 scores were even lower in the NAL than in the AFL (Table 2). This might indicate an even smaller ROM in the NAL; however, other features like the PC2 vector (amount of plantarflexion at push off) also changed the ROM.

With the PCA, the influence of the length of the stance phase (PC1 vector) on the vertical GRF could be separated from other changes in the waveforms. Contrary to the stance phase percentage, the difference in the PC1 scores was significant between the NAL and the controls. This suggests that the PC1 captures more than just the length of the stance phase, such as the slope of the GRF decrease at push-off. The PC2 vector of the GRF changed the amplitude, with the AFL of the patients having a lower amplitude than the controls (Table 2). Zeni Jr and Higginson [24] suggested that alterations in the peak values of the vertical GRF may arise as a result of altered walking speed. In our study, however, features that affect the peak values of the vertical GRF showed only significant changes for the AFL but not the NAL of asymmetric ankle OA patients. Therefore, physiological control of the vertical GRF may depend on more variables than walking speed alone. Clinically, this can be observed in the

limping gait pattern of asymmetric ankle OA patients who place the AFL subtly and slowly at heel strike to prevent painful high and shear joint loading. Subsequently, after midstance the NAL is loaded as fast as possible to reduce joint loading of the AFL during stance phase. The fast transition from loading the AFL to the NAL may be responsible for the higher vertical GRF of the NAL despite the same walking speed. Additionally, deficits in muscle activation of the AFL could also be a reason for lower push-off forces and thus lower GRFs during walking. These differences in the PCA results for the AFL and the NAL correspond with earlier findings that showed that pain has more influence on the walking ability (walking speed, cadence, peak GRF, etc.) than the ROM [5,8,25].

PC scores were used in a linear SVM classifier, which proved to be able to discriminate between the AFL of the patients and the non-dominant leg of the controls with a high recognition rate of 97.8%. This confirms that the data reduction using a PCA resulted in features with characteristic differences in the gait pattern of healthy subjects and asymmetric ankle OA patients. The robustness of the classification has now to be tested in a clinical environment on a larger group of subjects.

This study has a few limitations such as a small number of asymmetric ankle OA patients, differences in the number of males and females per group, and possible influence of significant differences in body weight. In order to have a perfectly clean group of asymmetric ankle OA patients, hard inclusion and exclusion criteria together with a low prevalence of ankle OA [26] led to only eight patients in the present study. The generalization of the results is therefore limited. However, our results showed significant differences between the gait patterns of patients and controls and even allowed us to classify between groups with a high recognition rate. The differences in the number of males and females per group as well as the differences in body weight might account for some of the differences in the gait patterns. But after the normalization of the kinetic gait data to body weight these differences are small and can be neglected.

In conclusion, our study showed that gait parameters such as walking speed and hindfoot ROM are reduced in patients suffering from asymmetric ankle OA. PCA allowed us to precisely quantify temporal changes in the waveforms of the hindfoot dorsiflexion angle and the vertical GRF. In contrast to the conventional analysis, PCA didn’t need any a priori decision on the parameter selection within a waveform. Additionally, PC scores could successfully be used to discriminate asymmetric ankle OA gait patterns from healthy ones using a linear SVM classification.

Skilled and experienced observers are able to recognize certain movement aspects; however, such subtle differences are often difficult to quantify. The results of the present study indicate that the combination of PCA and SVM can help to detect the significant elements of the altered gait patterns. This work therefore represents a step towards a objective quantification of gait data which is less dependent of the human assessment. The proposed quantification in this study should in the future be tested on a larger sample size to validate the results and improve the generalization. Furthermore, future studies could incorporate a longitudinal

design that may allow us to discern which variables contribute to disease progression and which may improve through joint preserving surgery aiming to delay progression of asymmetric ankle joint degeneration.

Acknowledgment:

This study was supported by the Swiss National Science Foundation (SNF No. 3200BO-120631). The funding source played no role in the study design, in the collection, analysis, or interpretation of the data, or in the writing of the manuscript or decision to submit the manuscript for publication. The authors also thank Beat Goepfert for his technical support.

Conflict of interest statement:

There is no conflict of interest.

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Chapter 3

Muscle activation of patients suffering from asymmetric ankle osteoarthritis during isometric contractions and level walking - a time-frequency analysis

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An adapted version was published in: Journal of Electromyography and Kinesiology. 2012; 22(6):939-946

Asymmetric osteoarthritis (OA) is a common type of OA in the ankle joint. OA also influences the muscles surrounding a joint, however, little is known about the muscle activation in asymmetric ankle OA. Therefore, the aim of this study was to characterize the patients' muscle activation during isometric ankle torque measurements and level walking. Surface electromyography (EMG) was measured of *Mm. gastrocnemius medialis* (GM) and *lateralis* (GL), *soleus* (SO), *tibialis anterior* (TA), and *peroneus longus* (PL) in 12 healthy subjects and 12 ankle OA patients. To obtain time and frequency components of the EMG power a wavelet transformation was performed. Furthermore, entropy was introduced to characterize the homogeneity of the wavelet patterns.
Patients produced lower plantar- and dorsiflexion torques and their TA wavelet spectra were shifted towards lower frequencies. While walking, the patients' muscles were active with a lower intensity and over a broader time-frequency region. In contrast to controls and varus OA patients, maximal GM activity of valgus OA patients lagged behind the activity of GL and SO. In both tasks, PL of the valgus patients contained more low frequency power. The results of this study will help to assess whether surgical interventions of ankle OA can reestablish the muscle activation patterns.

Keywords: ankle osteoarthritis; electromyography; wavelet analysis; entropy

1. Introduction

Osteoarthritis (OA) of the ankle joint typically develops after trauma and appears often asymmetrically, i.e. partially degenerated joint surface and joint axis deviation into varus or valgus. Asymmetric ankle OA is characterized by a progressing degeneration of articular cartilage, pain, and limitations of the joint mobility. In addition, OA also affects the muscles surrounding the joint. Several studies detected muscle weakness and muscle atrophy in patients suffering from OA in different joints [1,2]. This is believed to be caused by a reduced physical activity level and by arthrogenous muscle inhibition [3,4]. Muscle atrophy leads to changes in electromyographic (EMG) signals, characterized by changes in amplitude and frequency. In ankle OA, muscle atrophy resulted in lower ankle torques, a lower mean EMG frequency, and a lower mean intensity during maximal isometric contractions [1]. Findings from muscular activation during postural tasks in knee OA patients suggest that a lower mean frequency in OA is not only present in isometric contractions but also in dynamic ones with small joint movements [5].

Earlier stages of asymmetric ankle OA are often asymmetric. It is known, that the OA severity affects the characteristics of the muscle activation during walking. Despite a similar walking speed, the timing and intensity of the muscle activation differed between controls and patients with moderate or severe knee OA [6]. Moderate ankle OA patients have, on average, less pain and better functional scores than patients with severe end-stage ankle OA [7]. Their muscle weakness might therefore be less pronounced and different changes in the EMG signal than in end-stage ankle OA might be present.

To date, little is known about the muscle activation in patients suffering from ankle OA and even less about the spectral properties of the dynamic EMG signals derived during walking. To extract the properties of the EMG signal in a way that has been proven to be sensitive to differences caused by OA, the wavelet transformation analysis method of EMG signals was used [8,9]. The advantage of the wavelet transformation is that it doesn't require stationary signals and hence, is suited for dynamic signals. It resolves the power of the EMG signals in time and frequency with a resolution in both domains that allows observing physiologically relevant changes. The resulting wavelet patterns can depict the interplay between low and high intensity components of EMG signals and were used to investigate the muscle activation of end-stage ankle OA patients during walking [9]. The study demonstrated that the wavelet patterns of the patients had distinct changes that could be used for classification. However, due to the non-linearity of the classifier it was not possible to tell whether the timing of the muscle activation or the shift in frequency of the EMG signals allowed for classification.

The aim of this study was to characterize muscle activation of asymmetric ankle OA patients during isometric contractions and level walking in both time and frequency domain by using wavelet transformation. To analyze the differences in the muscle activation, different parameters on timing, intensity distribution, and frequency distribution were extracted from the wavelet patterns. Additionally, the influence of the varus/valgus alignment of the hindfoot was investigated. The hypotheses of the present study were: (i) asymmetric ankle OA patients

produce lower dorsi- and plantarflexion torques, (ii) for both tasks, the EMG signals of the lower leg muscles are shifted towards lower frequencies in the patients, and (iii) characteristics of the patients' wavelet patterns show significant differences to the controls.

2. Methods

2.1 Subjects

The characteristics of the 24 subjects that participated in this study are summarized in Table 1. The 12 ankle OA patients (6 male, 6 female) were recruited from our orthopaedic outpatient clinic (4 valgus, 8 varus). They all suffered from posttraumatic unilateral asymmetric ankle OA. Exclusion criteria were diabetes mellitus, neurological disorders, full end-stage ankle OA, early unilateral ankle OA with symmetric alignment pattern, and asymmetric ankle OA with pathologies or posttraumatic entities of the contralateral leg. 12 healthy adults (7 male, 5 female), who didn't have any pain and had no history of surgeries on the lower extremities, formed the control group. The local ethics committee approved the study and all subjects signed informed consent before participation.

Table 1: Subject characteristics

	Controls (*n*=12)		**Patients (*n*=12)**		**Varus (*n*=8)**		**Valgus (*n*=4)**		***P*** [a]	***P*** [b]	***P*** [c]
	Mean	**SEM**	**Mean**	**SEM**	**Mean**	**SEM**	**Mean**	**SEM**			
Age (years)	48.41	3.12	56.60	3.34	53.60	3.93	63.34	5.53	0.135	0.374	0.078
Height (m)	1.72	0.04	1.75	0.02	1.73	0.03	1.79	0.03	0.683	1.000	0.331
Weight (kg)	73.33	3.64	86.62*	4.16	86.74*	5.32	86.33	7.47	0.018	0.025	0.170
BMI (kg/m^2)	24.69	0.91	28.09*	0.96	28.66*	1.20	26.79	1.58	0.028	0.021	0.379
Pain (VAS)	0.00	0.00	5.31*	0.54	5.17*	0.75	5.63*	0.63	< 0.001	< 0.001	0.001
AOFAS score	100.00	0.00	61.38*	3.95	64.00*	3.73	55.50*	10.12	< 0.001	< 0.001	0.001
CC, NAF (cm)	38.29	0.77	39.42	0.79	40.33	0.88	37.38	1.21	0.326	0.101	0.579
CC, AFF (cm)	38.17	0.89	38.19	0.68	39.00	0.83	36.38	0.55	0.935	0.433	0.276
Difference CC (cm)	0.13	0.16	1.23*	0.41	1.33*	0.53	1.00	0.71	0.020	0.018	0.300

VAS: Visual Analogue Scale (0: no pain, 10: maximal pain); AOFAS: American Orthopaedic Foot and Ankle Score [10]; CC: maximal calf circumference; NAF: non-affected leg of the patients or dominant leg of the controls; AFF: affected leg of the patients or non-dominant leg of the controls; SEM: standard error of the mean

*: significant difference (*P*-value < 0.05) between controls and patients

[a]: P-value of Wilcoxon ranksum test between controls and all patients

[b]: P-value of Wilcoxon ranksum test between controls and varus patients

[c]: P-value of Wilcoxon ranksum test between controls and valgus patients

2.2 Data recording

The subjects completed six walking trials at self-selected speed on a laboratory walkway with two embedded force plates (Kistler, Winterthur, Switzerland). A six camera Vicon system (Vicon MX, Oxford, UK) recorded the positions of reflective markers that were placed on anatomical landmarks according to the Vicon plug-in gait model [11] for the calculation of the ankle moment and power. Bilateral EMG signals (recording system: Biovision, Wehrheim, Germany. Bandwidth 10–700 Hz, gain range 1000–5000) from *Mm. gastrocnemius medialis* (GM), *gastrocnemius lateralis* (GL), *soleus* (SO), *peroneus longus* (PL), and *tibialis anterior*

(TA) were recorded simultaneously at a sampling rate of 2400 Hz. The bipolar round Ag/AgCl surface electrodes (Noraxon U.S.A. Inc., Scottsdale, AZ, USA: 10 mm diameter, 22 mm interelectrode distance) were placed according to the European recommendations for surface electromyography (SENIAM) [12], after shaving and cleaning the skin with alcohol. To reduce movement artifacts, amplifiers and cables were taped to the skin. The ground electrode was placed over the tibial tubercle.

After the walking trials the subjects performed an isometric plantarflexion and dorsiflexion torque measurement with simultaneous EMG recordings at a sampling frequency of 2400 Hz [13]. The subjects were asked to perform maximal voluntary contractions lasting for five seconds. Five trials for each side and movement direction were recorded, with a break of two minutes between trials. The order of the exercises was chosen randomly.

2.3 EMG data processing

For the walking trials, the EMG signals were cut to one gait cycle per trial of each leg (heel strike to next ipsilateral heel strike) according to the position of the heel marker. In the isometric torque measurements the first 2 seconds after the onset of the torque plateau were analyzed as described below.

All EMG signals were analyzed in time-frequency space with a wavelet transformation based on 13 non-linearly scaled wavelets [8]. The wavelet transformation decomposed the power of the EMG signal in time and frequency components. The wavelets were indexed by j (w_j) and characterized by center frequency (Hz), bandwidth (Hz) and time resolution (ms): w_0 (7 Hz, 12 Hz, 80 ms); w_1 (19 Hz, 22 Hz, 53 ms); w_2 (38 Hz, 30 Hz, 39 ms); w_3 (62 Hz, 39 Hz, 30 ms); w_4 (92 Hz, 47 Hz, 25 ms); w_5 (128 Hz, 59 Hz, 21 ms); w_6 (170 Hz, 66 Hz, 19 ms); w_7 (218 Hz, 76 Hz, 17 ms); w_8 (271 Hz, 84 Hz, 15 ms); w_9 (331 Hz, 94 Hz, 13 ms); w_{10} (395 Hz, 101 Hz, 13 ms); w_{11} (466 Hz, 111 Hz, 11 ms); w_{12} (542 Hz, 118 Hz, 11 ms). In order to allow a comparison of the signals independent of the walking speed, a time normalization of the power was performed using a resolution of 1/600 of a gait cycle [9]. The EMG signals from the isometric contraction were wavelet transformed and resampled to 500 Hz.

The lowest wavelet w_0 was omitted from the analysis as it is highly influenced by movement artefacts [14]. High-frequency noise (at 400 Hz and higher) was present in some subjects' signals, thus only wavelets w_1 (19 Hz) to w_{10} (395 Hz) were further used. For walking, the results of the wavelet transformation could therefore be depicted by a wavelet pattern (matrix *WP*: 601 time points x 10 wavelets), with the abscissa representing the time in percent of the gait cycle, the ordinate the frequencies, and the grayscale the intensities. By summing the intensities at each time point, one receives the total intensity, and by summing the intensities at each frequency, one receives the wavelet spectrum. Since the interest in the distribution of the energy of the signal in time and frequency the wavelet patterns were normalized to the total energy (sum of all intensities).

To characterize the homogeneity of the muscle activation in the time-frequency domain, the entropy of each individual wavelet pattern was calculated according to the following formula by Shannon [15] adapted to the matrix *WP*:

$$E = -\sum_{j=1}^{10}\sum_{t=1}^{601} WP(t, j) * \log_2(WP(t, j))$$

WP(t,j): Intensity of the wavelet pattern at time *t* and frequency *j*

The entropy is maximal when all intensities are equally probable, thus $E = \log_2 6010 = 12.553$ and is equal to zero when all intensities except one have a probability of zero. The time of the maximal muscle activity during walking was obtained by the time of the maximal total intensity. To compare the distribution of the energy over the different frequencies, the wavelet pattern was divided into four frequency regions: w_1 (19 Hz) to w_3 (62 Hz), w_4 (92 Hz) to w_5 (128 Hz), w_6 (170 Hz) to w_8 (271 Hz), and w_9 (331 Hz) to w_{10} (395 Hz).

2.4 Statistics

All data are presented as mean ± standard error of the mean (SEM). For the analysis, only data from the non-dominant leg of the controls and the affected leg of the patients were used. Leg dominance was assessed by determining the preferred hopping leg [16]. A Wilcoxon rank sum test was used to test for statistical significance between the groups. The significance level was set at $\alpha = 0.05$.

3. Results

3.1 Isometric torque measurement

The patients produced a significantly lower plantarflexion and dorsiflexion torques with their affected leg than the controls. With regard to the hindfoot alignment a similar result was seen, although not always significant (Table 2).

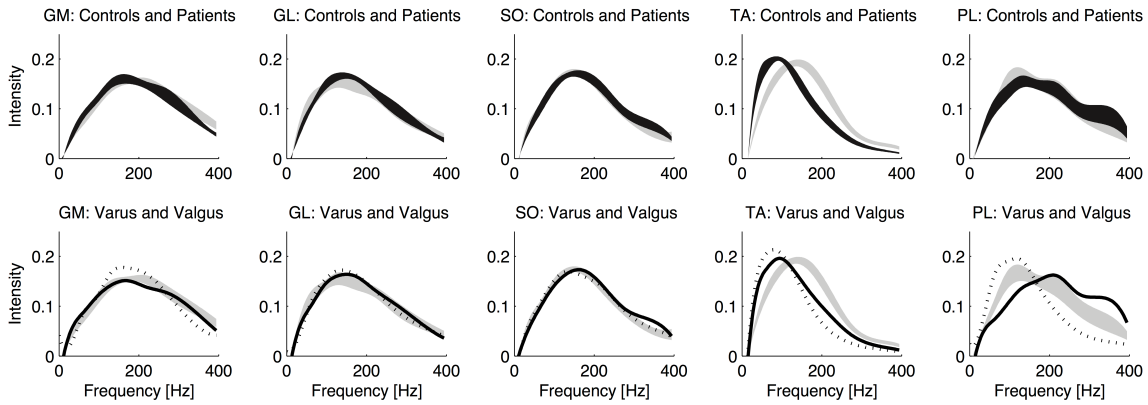

Figure 1: Mean wavelet spectra of the patients and controls. The upper panel shows the mean ± SEM of the controls (grey) and all patients (black), whereas the lower panel shows the mean ± SEM of the controls (grey), as well as the averages of the varus (solid) and valgus ankle OA patients (dotted). Data were interpolated using a cubic spline.

While there were no differences in the wavelet spectra of GM, GL, PL, and SO, a shift towards lower frequencies was detected in the mean wavelet spectrum of TA in patients compared to controls (Figure 1, upper row). In the valgus patients, the wavelet spectrum of PL showed a shift towards lower frequencies compared to the varus patients (Figure 1, lower row).

Table 2: Gait characteristics and maximal isometric ankle torques

	Controls (*n*=12)		Patients (*n*=12)		Varus (*n*=8)		Valgus (*n*=4)		P^a	P^b	P^c
	Mean	**SEM**	**Mean**	**SEM**	**Mean**	**SEM**	**Mean**	**SEM**			
Walking Speed (m/s)	1.25	0.01	0.98*	0.02	0.98*	0.03	0.97	0.02	0.014	0.041	0.058
Cadence (steps/min)	111.62	0.13	103.32*	0.52	104.58	0.72	100.81*	0.70	0.026	0.132	0.002
Stance phase (%)	60.74	0.06	62.15	0.11	62.05	0.11	62.34	0.28	0.285	0.263	0.684
Stride time (s)	1.08	0.00	1.17*	0.01	1.16	0.01	1.19*	0.01	0.026	0.132	0.002
Stride length (m)	1.35	0.01	1.13	0.03	1.13	0.04	1.14	0.02	0.078	0.153	0.170
Plantarflexion moment (Nm/kg)	1.78	0.01	1.30*	0.06	1.49*	0.07	0.92*	0.13	0.004	0.034	0.002
Ankle joint power (W/kg)	4.78	0.04	2.97*	0.09	3.56	0.12	1.77*	0.09	0.012	0.097	0.004
Maximal isometric plantarflexion torque (Nm)	36.9	3.9	20.9*	2.2	20.1*	2.5	22.6*	5.0	0.001	0.002	0.030
Maximal isometric dorsiflexion torque (Nm)	26.3	4.0	11.1*	2.3	11.8*	2.9	9.5	4.0	0.007	0.019	0.058

SEM: standard error of the mean
*: significant difference (*P*-value < 0.05) between controls and patients
[a]: *P*-value of Wilcoxon ranksum test between controls and all patients
[b]: *P*-value of Wilcoxon ranksum test between controls and varus patients
[c]: *P*-value of Wilcoxon ranksum test between controls and valgus patients

3.2 Gait

While the ankle OA patients walked significantly slower and with a lower cadence than the healthy controls, the differences in the length of the stance phase and the stride length were not statistically significant (Table 2). Additionally, patients produced a significantly lower peak plantarflexion moment and peak ankle power during walking. However, taking into account the hindfoot alignment, not all of these differences were statistically significant (Table 2).

3.2.1 Wavelet EMG pattern

The mean wavelet patterns during a gait cycle of the non-dominant leg of the controls and the affected leg of the patients are depicted in Figure 2. As the grey scale is the same for each muscle, one can see that on average the intensity was lower in the patients compared to the controls. Additionally, the main area of muscle activation in the patients was broader and less distinct than in the controls. For SO and PL, this resulted in a significantly higher entropy in the patients (SO: 10.53 ± 0.10; PL: 10.76 ± 0.12) than in the controls (SO: 10.10 ± 0.12; PL: 10.17 ± 0.14) (SO: $p = 0.014$; PL: $p = 0.009$).

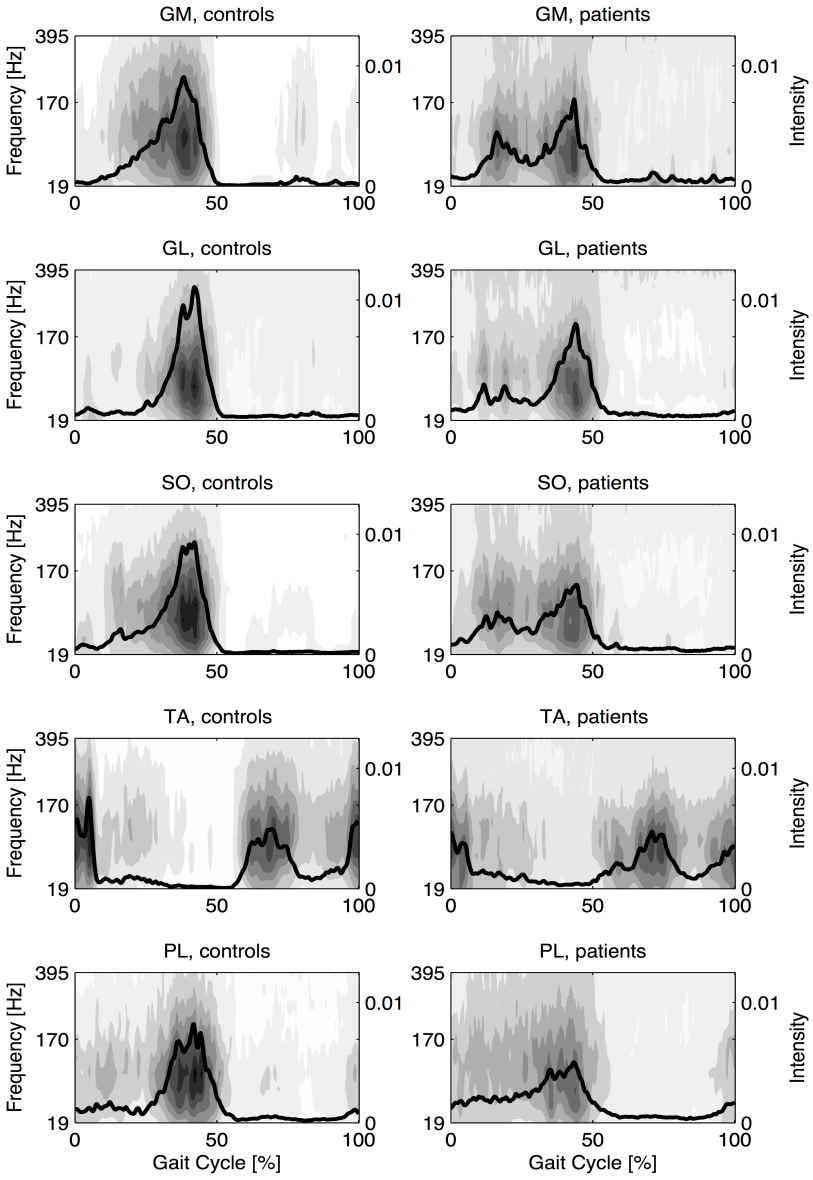

Figure 2: Mean wavelet patterns and total intensities (black line) of the different lower leg muscles during one gait cycle. The darker colors indicate higher intensities and thus higher muscle activity. Left panel: non-dominant leg of the controls; right panel: affected leg of the patients.

3.2.2 Muscle activation in the time domain

As Figure 2 shows the plantarflexion muscles (GM, GL, SO, PL) were mainly active in the late stance phase whereas TA was mainly active during the early swing phase and around heel strike. The maximal intensity occurred between 34.7 and 40.1% of the gait cycle in the plantarflexors and between 67.7 and 69.3% in TA (Figure 3) with no significant differences between the groups. The visual inspection of the mean wavelet patterns and total intensities revealed that the muscle activation of TA started earlier in the patients than in the controls. Furthermore, patients showed an additional peak in GM and SO during the early stance phase (Figure 2).

The total intensity in the mean wavelet patterns also indicated a timing difference of the maximal muscle activation of the triceps surae (GM, GL, and SO). While this difference was negative for GM-GL and GM-SO in the controls, it was smaller and positive in the patients. The timing difference in GL-SO was close to zero for both groups indicating that the peak intensities occurred at the same time (Figure 3).

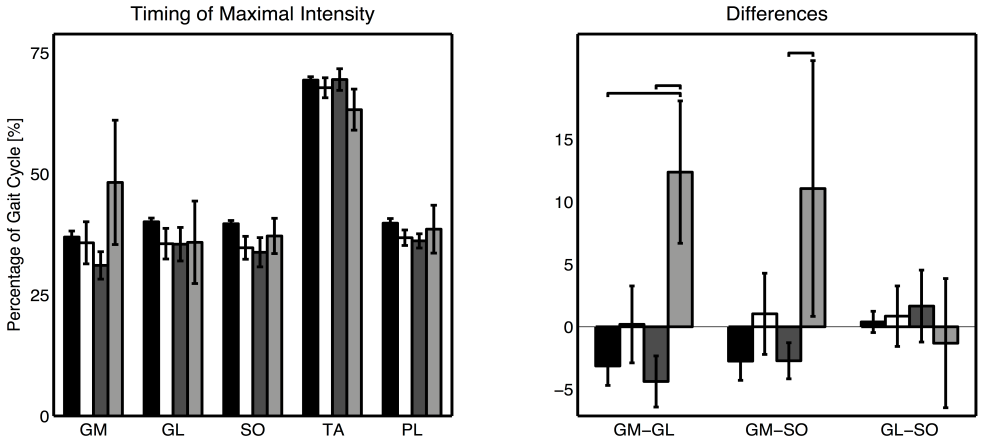

Figure 3: Time of maximal intensity (left) and differences between the times of maximal intensity (right). Mean ± SEM of the controls (black), all patients (white), the patients with varus ankle OA (dark grey), and the patients with valgus ankle OA (light grey). Horizontal lines indicate significant group differences.

3.2.3 Frequency distribution

The wavelet patterns of the patients had less energy between 92 and 128 Hz than controls for TA (p = 0.019). Contrary to the controls, the wavelet patterns of the patients contained more frequency components in the range of 331 to 395 Hz for SO (p = 0.012) and PL (p = 0.010). Additionally, there were trends that in patients TA had more energy between 19 and 62 Hz (p = 0.069) and less energy between 170 and 271 Hz (p = 0.061), GM less energy between 92 and 128 Hz (p = 0.078) and between 170 and 271 Hz (p = 0.089), and PL less energy between 92 and 128 Hz (p = 0.078) (Figure 4).

3.2.4 Differences between varus and valgus ankle OA

Differences in the muscle activation patterns between varus and valgus patients were seen in PL (Figure 5). The PL wavelet patterns of the varus patients contained significantly less energy between 19 and 62 Hz (p = 0.016) and more energy between 170 and 271 Hz (p = 0.0108) than the valgus patients (Figure 4). Furthermore, a significant difference in the timing of the peak intensity within the triceps surae muscles (GM, GL, and SO) was seen. While in varus patients and controls, GM was maximally active earlier than both GL and SO, the opposite was seen in valgus patients. The difference between the peak intensity locations GM-SO was significantly higher in valgus patients than in varus patients (p = 0.048) and higher than in controls (p = 0.070), while GM-GL was significantly higher in valgus than in both varus (p = 0.024) and controls (p = 0.004) (Figure 3).

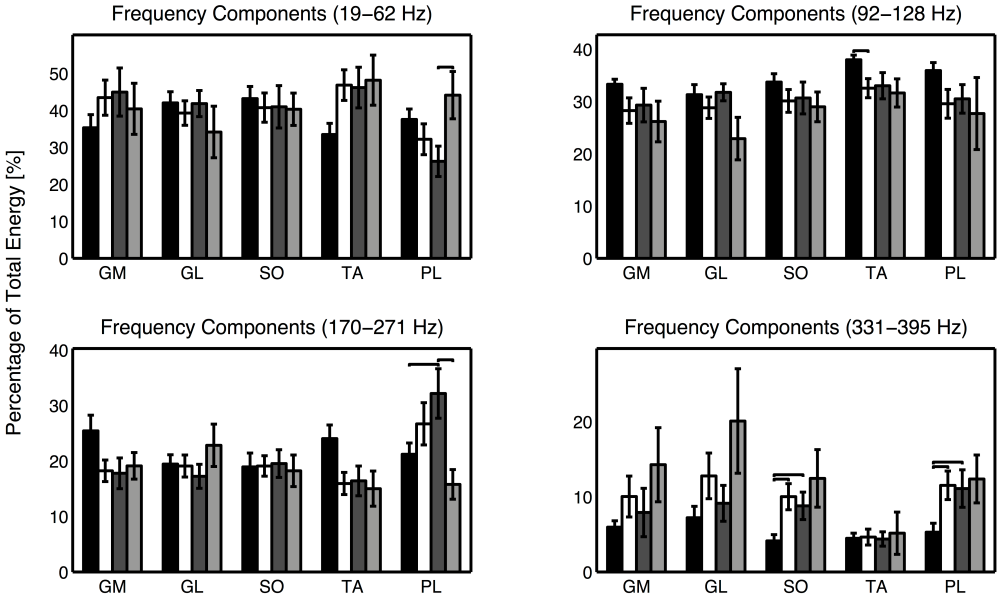

Figure 4: Distribution of the total energy over different frequency regions. Mean ± SEM of the controls (black), all patients (white), the patients with varus ankle OA (dark grey), and the patients with valgus ankle OA (light grey). Horizontal lines indicate significant group differences.

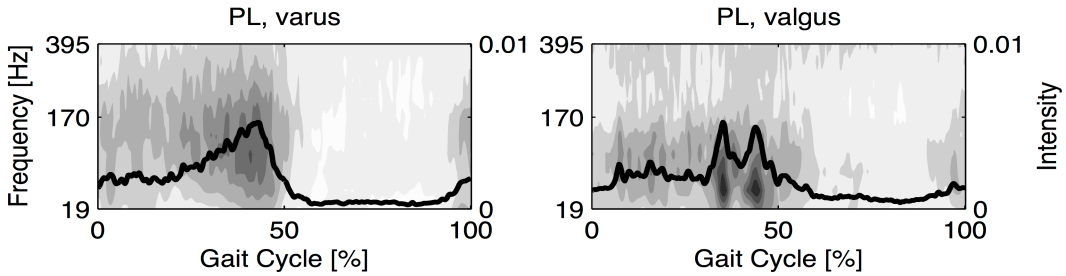

Figure 5: Mean wavelet patterns and total intensities (black line) of the patients' *M. peroneus longus* (PL) during one gait cycle. Left panel: mean of patients suffering from varus ankle OA; right panel: mean of patients suffering from valgus ankle OA.

4. Discussion

Patients with asymmetric ankle OA produced lower isometric torques, walked slower and had lower ankle joint moments during gait. These results are in agreement with previous literature on end-stage ankle OA [17,18] and confirm our first hypothesis of lower force levels in asymmetric ankle OA. This means that although the OA of the patients in this study was less severe, their overall gait pattern was affected in a similar way as in end-stage full OA.

In both tasks, maximal isometric contraction and walking, patients produced less force in plantarflexion. Although the walking speed was comparable, the valgus patients produced a lower plantarflexion moment than the varus patients. The lower ankle joint moment was also

reflected in a lower intensity in the mean wavelet patterns during walking. However, the temporal activation of the main plantarflexion muscles only showed small differences. In valgus ankle OA patients GL and SO were maximally active before GM, whereas GM was active earlier than GL and SO in varus ankle OA patients and healthy controls. This indicates that the inter-muscular coordination during push-off was altered in valgus ankle OA patients. During push-off, the hindfoot normally rotates into a varus position and locks the midfoot to stabilize the foot [19]. It seems that in valgus ankle OA patients this normal varisation of the hindfoot is reduced or even missing, thus changing the muscle activation and possibly the ankle joint moment. However, it remains unclear whether the temporal changes in the activation of the triceps surae are a consequence of the valgus ankle OA or whether they play a role in the etiological development of valgus ankle OA. It also has to be noted that since there were only four patients in the valgus group the significance of these results is limited.

Contrary to our hypothesis of a shift towards lower frequencies in the wavelet spectra of patients and previous results on end-stage ankle OA [1], the spectra measured during the isometric contraction remained unchanged in GM, GL, and SO. Similarly, the energy distribution in the walking wavelet patterns of the patients resembled the ones from the controls, except for SO that had more energy in the range of 331 to 395 Hz. In contrast to end-stage ankle OA patients [1], the patients of the present study suffered from moderate ankle OA with only a slight reduction in calf circumference (1 cm) and higher functional scores. It is therefore possible that their lower functional limitation preserved the high-frequency components of the EMG signals in the calf muscles.

In contrast to the calf muscles that work concentrically and produce high forces during walking, TA is acting both eccentrically (early stance) and concentrically (swing phase) with low force productions [20]. Despite these functional differences, the hypothesis of a frequency shift was confirmed for TA in both tasks. The wavelet spectra of the torque measurements were shifted towards lower frequencies and the wavelet pattern during walking showed that in patients more energy was located in the lower frequencies (Figure 4). As suggested in previous literature on OA and spectral properties of EMG signals [1,5] the spectral changes in TA could be associated with a selective atrophy of type II muscle fibers. However, this would have to be confirmed by histological data.

Entropy has been previously used to study the spatial distribution of EMG signals [21]. In the present study, we used the entropy for the first time to quantify the homogeneity of the wavelet patterns. As indicated by the higher entropy in SO and PL of the patients, the wavelet patterns of these muscles had a less accented peak. It is therefore possible that the healthy controls had a more precise muscle activation and that the patients had to maintain their muscle activation over a longer time to increase the joint stability. Another possible interpretation is that the patients' muscle weakness and lower total intensity led to the broader muscle activation. The measure of entropy is ideally suited to assess the broader muscle activation and is likely a new variable that should be used in addition to the classical ones used in the EMG analysis. However, the entropy doesn't yield information on the location of the peaks within a wavelet pattern but together with the timing of the maximal intensity

(Figure 3) and the distribution of the total energy over different frequency regions (Figure 4), the variables supported the hypothesis of the detection of significant differences in the wavelet patterns.

The main differences between varus and valgus ankle OA patients were seen in PL. They were likely related to the asymmetry of the ankle OA and the malaligned ankle joint. Murley et al. [22] showed that the foot posture influences the muscle activity of PL. Subjects with flat-arched feet, which are related to hindfoot valgus, have lower PL activity in the second part of the stance phase. It was also seen that the mean EMG frequency increases with increased force levels [23]. Since the EMG signals were normalized to total energy in this study, the intensity of the activity of PL couldn't be compared between the different groups. However, it is possible that the lack of PL activity in valgus ankle OA was related to the higher amount of low-frequency components (19-62 Hz). Similarly, varus ankle OA required more PL activity to stabilize the ankle joint which in turn could explain the higher amount of high-frequency components in varus ankle OA patients. These changes in PL activation however, didn't result in changes of the torques or gait parameters, similar to the outcome of a recent study on ankle alignment and gait parameters in ankle OA patients [24].

There were some limitations in this study. As ankle OA is less prevalent than knee or hip OA [25], only a small number of patients met the clean inclusion criteria and could be included in the study. However, the observed changes in the muscle activation patterns were statistically significant. Additionally, the asymmetry of the ankle OA led to a rather inhomogeneous patient group that could have influenced the results, especially for the activation of PL. To further investigate the muscle activation with respect to hindfoot alignment, more patients in these two groups would be needed.
There are several factors that could influence the EMG signals and their interpretation. Fatigue, for example, leads to a shift towards lower frequencies in the EMG spectra. Since the patients only walked short distances with breaks in between and they were given sufficient rest between torque measurement trials, it is unlikely that the patients experienced fatigue.

In conclusion, this study showed that patients suffering from asymmetric ankle OA have altered muscle activation patterns. While changes in the frequency domain were observed independently of the task, the temporal activation of the lower leg muscles was influenced by the patients' hindfoot alignment. Measurements of the muscle activation during functional tasks such as walking and the documentation of the hindfoot axis (valgus, neutral, varus) are therefore needed in biomechanical ankle OA studies to analyze changes due to OA. They will further allow us to control whether surgical interventions can reestablish the muscle activation patterns of healthy subjects. Unhealthy muscle activation can cause stress on the joints and worsen the OA development. Measuring the muscle activation properties will therefore allow us to further investigate the influence of a deteriorating muscle activity. Contrary to previous results in ankle OA patients, this study showed no significant changes in the frequency content of the EMG signals, except for TA. A change in frequency of the EMG signal was mainly interpreted as a change in fiber type composition. This could therefore be a sign that the fiber type composition in earlier stages of ankle OA is less affected. However, this kind of

interpretation is still debated [26,27] and thus histological studies need to be done. Further studies are needed to investigate the recovery potential after muscle atrophy, especially with regard to fiber type distribution. One will also have to investigate the recovery potential or the preservation of the calf muscles' activation patterns, after early surgical interventions of ankle OA, such as joint preserving surgery.

Acknowledgment:

This study was supported by the Swiss National Science Foundation (SNF No. 3200BO-120631). The authors also thank Beat Goepfert for technical support and Philippe C. Cattin for help with the data analysis.

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Chapter 4

Mid- to long-term outcomes after realignment surgery in asymmetric ankle osteoarthritis: clinical and gait analysis

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Submitted to: Journal of Bone and Joint Surgery, American Volume

Background: Supramalleolar osteotomies have gained popularity for the treatment of early and mid-stage asymmetric ankle osteoarthritis. The present study analyzed spatiotemporal, kinematic and kinetic gait parameters, and functional outcomes in patients who underwent realignment surgery and in patients with asymmetric ankle osteoarthritis.
Methods: This study included three groups: patients with long-term follow-up after realignment surgery (n = 8), patients suffering from asymmetric ankle osteoarthritis (n = 8), and healthy controls (n = 8). Three-dimensional instrumented gait analysis was performed to assess spatiotemporal parameters, three-dimensional joint angles, ground reaction forces, and joint moments and powers. Clinical evaluation consisted of documentation of the pain level, range of motion, the American Orthopaedic Foot and Ankle Society (AOFAS) hindfoot score, the calf circumference, and SF-36 life quality score.
Results: Gait analysis demonstrated that the patients with ankle osteoarthritis had reduced cadence and peak ankle dorsiflexion moment, and prolonged stride and step time in comparison to control subjects and patients after realignment surgery. Hindfoot dorsiflexion, peak ankle power, and the second peak of the vertical ground reaction force were significantly lower in patients with ankle osteoarthritis and after realignment surgery as compared to controls. The realignment patients had significantly less pain and higher AOFAS hindfoot and SF-36 scores than patients with arthritic ankles.
Conclusion: Realignment surgery in patients with asymmetric ankle osteoarthritis led to improvements of the gait pattern. Long-term functional outcome and quality of life were superior to those observed in patients with asymmetric ankle osteoarthritis. Supramalleolar osteotomies are a promising treatment options in patients with asymmetric ankle osteoarthritis.
Level of Evidence: Therapeutic Level III (case-control study)

Keywords: realignment surgery; supramalleolar osteotomy; comparative study; gait analysis; functional outcome

1. Introduction

Many patients with ankle osteoarthritis (OA) have a posttraumatic etiology [1,2]. Patients with posttraumatic ankle OA are often younger than patients with OA of the hip or knee [3]. More than half of all patients with ankle OA present with a substantial malalignment of the hindfoot, where varus deformity is more frequent [2,4].

The current standard surgical treatment for patients with advanced ankle OA include ankle arthrodesis[5] and total ankle replacement.[6] Ankle arthrodesis has been shown to provide substantial pain relief in mid-term [7-9], but this procedure may cause persistent alteration in gait and progression of degenerative changes in neighboring joints [10,11]. Total ankle replacement is an increasingly recommended alternative treatment to ankle arthrodesis in patients with end-stage ankle OA [12-14]. However, in the current literature few studies address the long-term outcome of this procedure [15,16].

In patients with asymmetric varus or valgus ankle OA, realignment surgery, including supramalleolar osteotomies, may preserve the ankle joint [17,18]. Realignment surgery has been shown to provide pain relief and functional improvement in patients with both varus [7-29] and valgus ankle OA [17-20,22,30]. While gait outcomes in patients who underwent ankle arthrodesis or total ankle replacement have been addressed in numerous studies [13,31-39], little is known about the influence of supramalleolar osteotomies.

Therefore, we analyzed clinical and biomechanical outcomes in patients with asymmetric ankle OA who underwent realignment surgery with supramalleolar osteotomies. The objectives of our study were to (i) determine the spatiotemporal, kinematic, and kinetic gait parameters in patients with asymmetric ankle OA and in patients who underwent realignment surgery; (ii) assess the pain level in patients; and (iii) assess the patients' mid- to long-term functional outcomes, including range of motion (ROM) and quality of life.

2. Patients and methods

The protocol of this cross-sectional comparative study was approved by the Ethics Committee of the University of Basel, Basel, Switzerland. The study was conducted in accordance with the Declaration of Helsinki and the Guidelines on Good Clinical Practice [40,41]. All participants provided informed written consent prior to surgery and study participation.

From 2001 to 2003, eight adult patients underwent realignment surgery due to symptomatic ankle valgus alignment. Exclusion criteria for realignment surgery included unmanageable joint instability, neurovascular disease, end stage arthritis, contralateral ankle OA, and substantial comorbidities (e.g. diabetes mellitus). There were five male and three female patients with a mean age (and standard deviation) of 36.3 ± 6.8 years (range: 24.5 to 44.9 years). Other demographic data are presented in Table 1. The mean follow-up was 8.4 ± 0.7 years (range, 7.6 to 9.3 years).

The eight patients who underwent realignment surgery were matched to eight patients suffering from unilateral symptomatic ankle valgus alignment and to eight healthy subjects without any pathologies or previous surgeries of the ankle/hindfoot. All groups were of similar age, gender, height, body mass, and body mass index (Table 1).

Table 1: Demographic data on study participants. Data are presented as mean (range).

Parameter	Patients after realignment surgery	Patients with asymmetric ankle OA	Control group	*P* value
Age[#] (years)	44.7 (33.5 to 54.1)	56.9 (36.3 to 66.9)	50.4 (39.5 to 69.2)	0.077[†]
Gender M:F	5:3	5:3	4:4	0.842[‡]
Height (m)	1.75 (1.59 to 1.98)	1.75 (1.58 to 1.84)	1.71 (1.52 to 1.92)	0.712[†]
Body mass (kg)	84.8 (60.2 to 143.9)	83.7 (54.2 to 102.3)	72.0 (55.5 to 95.6)	0.338[†]
BMI* (kg/m^2)	27.2 (22.8 to 36.7)	27.3 (21.7 to 31.0)	24.6 (22.2 to 28.7)	0.235[†]
Side right:left	4:4	7:1	6:2	0.244[‡]

[#]at time of clinical and biomechanical assessment
*BMI: body mass index
[†]using ANOVA (analysis of variance) test
[‡]using Chi-Square test

2.1 Radiographic assessment

Preoperative and postoperative weight-bearing radiographs of the foot and ankle were performed for all patients who underwent realignment surgery. Radiographic assessment was also performed in patients with asymmetric ankle OA. Hindfoot alignment was assessed using the following radiographic parameters: medial distal tibial angle[42,43] (normal value, 89° to 92°) and tibiotalar angle[44] (normal value, 91.5° ± 1.2°). Ankle OA was graded using classification according to Takakura et al [28]. Preoperative radiographic findings in patients who underwent realignment surgery were comparable to those observed in the patient group with ankle OA (Table 2).

Table 2: Radiographic results. Data are presented as mean (range).

Parameter	Patients after realignment surgery			Patients with asymmetric ankle OA	*P* value[#]
	preoperative	postoperative	*P* value		
MDTA[§]	99.3° (97° to 102°)	90.0° (88° to 93°)	0.008[†]	100.4° (98° to 103°)	0.232[$]
Tibiotalar angle	102.9° (96° to 109°)	91.6° (90° to 93°)	< 0.001[‡]	100.5° (95° to 108°)	0.347[$]
Ankle OA grade*	Grade 1: 1 Grade 2: 4 Grade 3: 3	Grade 1: 1 Grade 2: 5 Grade 3: 2	1.000[†]	Grade 1: 0 Grade 2: 5 Grade 3: 3	0.798[£]

[#]Patients with realignment surgery (preoperative values) vs. patients with asymmetric ankle OA
[§]MDTA: medial distal tibial angle
*according to Takakura et al [28]: Grade 0, parallel joint, no tibiotalar tilt and no signs of arthritis; Grade 1, parallel joint, no tibiotalar tilt but signs of subchondral sclerosis or osteophytes formation; Grade 2, tibiotalar tilt with varus or valgus alignment without subchondral bone contact; Grade 3, tibiotalar tilt with varus or valgus alignment with subchondral bone contact; and Grade 4, total joint loosening with total subchondral bone contact
[†]using Wilcoxon signed rank test
[‡]using paired t-test
[$]unpaired t-test
[£]using Mann-Whitney rank sum test

2.2 Surgical technique

Surgical correction was planned with use of anteroposterior weight-bearing ankle radiographs as described previously [45,46]. Operative correction was done according to a previous report [18]. In all patient a supramalleolar medial closing-wedge osteotomy was performed via the medial approach [18,30]. The osteotomy was performed using an oscillating saw with continuous water irrigation to avoid thermal damage. After removal of the bone wedge, the osteotomy was closed and secured using a rigid plate with locking screws. After performing the osteotomy, heel alignment was reassessed clinically. In patients with remaining pes planovalgus et abductus deformity a lateral lengthening osteotomy of the calcaneus was performed [47]. In patients with ligamentous instability, medial and/or lateral reconstruction of ligaments was performed [48,49]. Additional procedures are shown in Table 3. After surgery, partial weight-bearing in a stable walker (VACOped; OPED, Cham, Switzerland) was permitted. After eight weeks, full weight-bearing was allowed and a rehabilitation physiotherapy program was started.

Table 3: Demographic data and additional surgical procedures in eight patients who underwent realignment surgery.

Case	Demographic data			Additional surgical procedures		
	Age (years)*	Gender	BMI[#]	Anterior cheilectomy	Calcaneal osteotomy	Ligamentous reconstruction
1	39.1	female	30.1	yes	no	no
2	24.5	male	22.8	no	no	medial/lateral
3	33.8	male	36.7	no	no	no
4	42.2	male	27.5	no	no	medial/lateral
5	33.1	female	23.8	yes	no	no
6	31.4	male	25.4	yes	yes	no
7	41.3	female	24.0	yes	no	no
8	44.9	male	27.0	no	no	no

[#]BMI: body mass index in kg/m^2

*at time of surgery

2.3 Clinical assessment

Patients rated their pain on a visual analogue scale (VAS) ranging from 0 (no pain) to 10 (maximal pain) [50]. Ankle ROM was determined with a goniometer placed along the lateral border of the leg and foot. Goniometer measurements were performed in the weight-bearing position, comparable with the method described by Lindsjö et al [51]. In addition, the American Orthopaedic Foot & Ankle Society (AOFAS) hindfoot score was calculated [52]. All subjects completed the Short Form-36 (SF-36) questionnaire [53]. The maximal circumference of both legs was measured at the widest calf level in a standing position and the corresponding side difference was calculated [39].

2.4 Gait analysis

Three-dimensional gait analysis was performed using a six camera Vicon system (Vicon MX3+, Oxford, UK) with a sampling rate of 120 Hz. Two force plates (Kistler Instrument Corp., Amherst, NY, USA) embedded in the laboratory walkway measured the three-

dimensional ground reaction forces with a sampling rate of 2400 Hz. Reflective markers were placed on predefined anatomical landmarks according to the Oxford foot model [54] and the Plug-In Gait model [55]. The subjects were allowed several warm-up trials to get accustomed to the laboratory setting. A total of six barefoot walking trials were recorded at a comfortable, self-selected speed. Trials were repeated if the subjects didn't hit the force plates centrally.

Joint kinematics and kinetics were calculated by the Vicon Nexus software (Plug-In Gait and Oxford foot model, Vicon Motion Systems, Oxford, UK). For the three-dimensional joint angles, external joint moments, and ground reaction forces the minimal and maximal values, as well as ranges of motion were determined in each trial and then averaged within the subjects. Similarly, the spatiotemporal gait parameters were averaged.

2.5 Statistical analysis

Post-hoc power analyses indicated that the study was adequately powered at a level of $P > 0.80$ for eight subjects in each group. A Kolmogorov-Smirnov normality test was performed to verify that the data met the assumptions of a parametric test. An analysis of variance test and Chi-Square test were used for comparison of continuous and non-continuous variables, respectively. Post hoc comparisons between the three groups were performed with the use of the Tukey honestly significant difference (HSD) test. The following parametric and non-parametric tests were used to compare two groups: unpaired/paired t-test and Mann-Whitney rank sum test/Wilcoxon signed rank test. The level of significance was set at $p \leq 0.05$.

2.6 Source of funding

The authors thank the Swiss National Science Foundation (SNF No. 3200BO-120631) for financial support.

3. Results

3.1 Spatiotemporal parameters

The mean cadence in patients with asymmetric ankle OA was significantly lower than in the control group, while there was no difference between patients after realignment surgery and subjects in the control group. The mean walking speed, stride time, and step time in patients with asymmetric ankle OA were the lowest among the three groups. The length of the stance phase, stride length, and step length were comparable in all three groups (Table 4).

3.2 Kinematic and kinetic parameters

The hip joint ROM was comparable in all three groups. Statistically significant differences between groups were found for knee flexion, hindfoot dorsiflexion and rotation, forefoot supination, and metatarsophalangeal-I (MTP-I) dorsiflexion (Table 5). The temporal

progression of these angles during a gait cycle (except the hindfoot rotation) is depicted in Figure 1.

Table 4: Spatiotemporal gait parameters. Data are presented as mean ± standard deviation.

Parameter	Patients after realignment surgery	Patients with asymmetric ankle OA	Control group	*P* value
Cadence (steps/min)	107.3 ± 5.0	102.6 ± 6.3	114.3 ± 7.1	0.004[†] 0.292; 0.086; 0.003[‡]
Walking speed (m/s)	1.18 ± 0.08	1.09 ± 0.15	1.25 ± 0.09	0.038[†] 0.280; 0.468; 0.030[‡]
Stride time (s)	1.12 ± 0.05	1.17 ± 0.07	1.05 ± 0.07	0.004[†] 0.227; 0.112; 0.003[‡]
Step time (s)	0.57 ± 0.03	0.60 ± 0.05	0.53 ± 0.03	0.002[†] 0.200; 0.066; 0.001[‡]
Stance phase (% gait cycle)	60.0 ± 1.4	60.8 ± 1.6	60.4 ± 1.4	0.598[†]
Stride length (m)	1.32 ± 0.09	1.27 ± 0.12	1.31 ± 0.14	0.692[†]
Step length (m)	0.67 ± 0.04	0.65 ± 0.07	0.65 ± 0.07	0.724[†]

[†]using ANOVA (analysis of variance) test
[‡]using Tukey HSD (honestly significant difference) post hoc test: patients after realignment surgery vs. patients with asymmetric ankle OA; patients after realignment surgery vs. control group; patients with asymmetric ankle OA vs. control group

Table 5: Range of motion during gait. Data are presented as mean ± standard deviation.

Parameter	Patients after realignment surgery	Patients with asymmetric ankle OA	Control group	*P* value
Hip flexion (°)	45.9 ± 4.8	40.8 ± 6.7	43.6 ± 3.8	0.177[†]
Hip adduction (°)	11.9 ± 4.5	9.6 ± 1.9	12.5 ± 3.9	0.253[†]
Hip rotation (°)	25.4 ± 5.8	24.0 ± 3.4	26.6 ± 8.0	0.704[†]
Knee flexion (°)	61.8 ± 4.5	53.0 ± 6.5	59.3 ± 4.9	0.011[†] 0.010; 0.632; 0.072[‡]
Hindfoot dorsiflexion (°)	13.2 ± 3.4	17.6 ± 4.6	23.1 ± 4.2	< 0.001[†] 0.104; < 0.001; 0.035[‡]
Hindfoot inversion (°)	15.1 ± 3.5	14.7 ± 2.4	15.6 ± 5.6	0.901[†]
Hindfoot rotation (°)	8.9 ± 4.5	10.9 ± 4.4	14.1 ± 2.7	0.043[†] 0.545; 0.035; 0.258[‡]
Forefoot dorsiflexion (°)	14.6 ± 2.2	18.4 ± 7.4	16.6 ± 2.8	0.303[†]
Forefoot supination (°)	8.9 ± 2.2	9.1 ± 3.2	13.4 ± 3.7	0.013[†] 0.988; 0.022; 0.030[‡]
Forefoot rotation (°)	9.2 ± 1.7	9.2 ± 3.1	9.0 ± 1.7	0.990[†]
MTP-I[#] dorsiflexion (°)	19.0 ± 5.0	19.9 ± 7.5	28.0 ± 8.1	0.033[†] 0.958; 0.044; 0.077[‡]

[#]MTP-I: metatarsophalangeal-I joint
[†]using ANOVA (analysis of variance) test
[‡]using Tukey HSD (honestly significant difference) post hoc test: patients after realignment surgery vs. patients with asymmetric ankle OA; patients after realignment surgery vs. control group; patients with asymmetric ankle OA vs. control group

The temporal progression of the vertical ground reaction force, ankle dorsiflexion moment, and ankle power during a gait cycle is depicted for all three groups in Figure 2. Both patients after realignment surgery and patients with asymmetric ankle OA had a significantly lower ankle power than controls. For the peak ground reaction forces, the second vertical peak (during push off) was significantly lower in both patient groups, while the maximal anterior ground reaction force was the lowest in patients with asymmetric ankle OA. The kinetic parameters measured in the knee and hip joints were comparable in all three groups (Table 6).

Table 6: Kinetic parameters. Data are presented as mean ± standard deviation.

Parameter	Patients after realignment surgery	Patients with asymmetric ankle OA	Control group	*P* value
Maximal medial GRF (N/kg)	0.71 ± 0.20	0.63 ± 0.17	0.62 ± 0.13	0.511[†]
Maximal posterior GRF (breaking) (N/kg)	1.62 ± 0.33	1.48 ± 0.33	1.69 ± 0.27	0.411[†]
Maximal anterior GRF (propulsion) (N/kg)	1.77 ± 0.22	1.59 ± 0.53	2.10 ± 0.29	0.040[†] 0.602; 0.215 0.034[‡]
1st peak vertical GRF (loading response) (N/kg)	10.53 ± 0.29	10.52 ± 0.66	10.99 ± 0.35	0.086[†]
Trough vertical GRF (mid stance) (N/kg)	8.14 ± 0.52	8.25 ± 0.75	7.66 ± 0.64	0.173[†]
2nd peak vertical GRF (push-off) (N/kg)	10.57 ± 0.72	10.40 ± 1.15	11.96 ± 0.52	0.002[†] 0.917; 0.009; 0.004[‡]
Maximal hip flexion moment (Nm/kg)	1.11 ± 0.22	0.97 ± 0.29	1.09 ± 0.17	0.412[†]
Maximal hip extension moment (Nm/kg)	0.77 ± 0.16	0.68 ± 0.41	0.82 ± 0.28	0.651[†]
Maximal hip adduction moment (Nm/kg)	0.94 ± 0.22	1.04 ± 0.12	1.00 ± 0.20	0.539[†]
Maximal knee flexion moment (early stance phase) (Nm/kg)	0.44 ± 0.25	0.47 ± 0.28	0.46 ± 0.28	0.965[†]
Maximal knee extension moment (late stance phase) (Nm/kg)	0.42 ± 0.13	0.23 ± 0.19	0.38 ± 0.21	0.121[†]
Maximal ankle dorsiflexion moment (Nm/kg)	1.51 ± 0.18	1.41 ± 0.40	1.80 ± 0. 15	0.023[†] 0.766; 0.098; 0.023[‡]
Ankle power (W/kg)	2.93 ± 0.75	3.15 ± 1.69	4.86 ± 0.63	0.005[†] 0.918; 0.007; 0.017[‡]

[#]MTP-I: metatarsophalangeal-I joint

[†]using ANOVA (analysis of variance) test

[‡]using Tukey HSD (honestly significant difference) post hoc test: patients after realignment surgery vs. patients with asymmetric ankle OA; patients after realignment surgery vs. control group; patients with asymmetric ankle OA vs. control group

3.3 Clinical Results

While the dorsiflexion of the ankle joint was lower in both patient groups, the plantar flexion was only significantly lower in the patients after realignment surgery compared to the control group. The maximal calf circumference was comparable in all three groups. Patients with asymmetric ankle OA had significantly lower SF-36 and AOFAS hindfoot scores and higher VAS scores than patients after realignment surgery and the control group. Results were generally better in patients after realignment surgery than in patients with asymmetric ankle OA. However, for pain level and AOFAS hindfoot score there was a statistically significant difference between the affected groups and controls (Table 7).

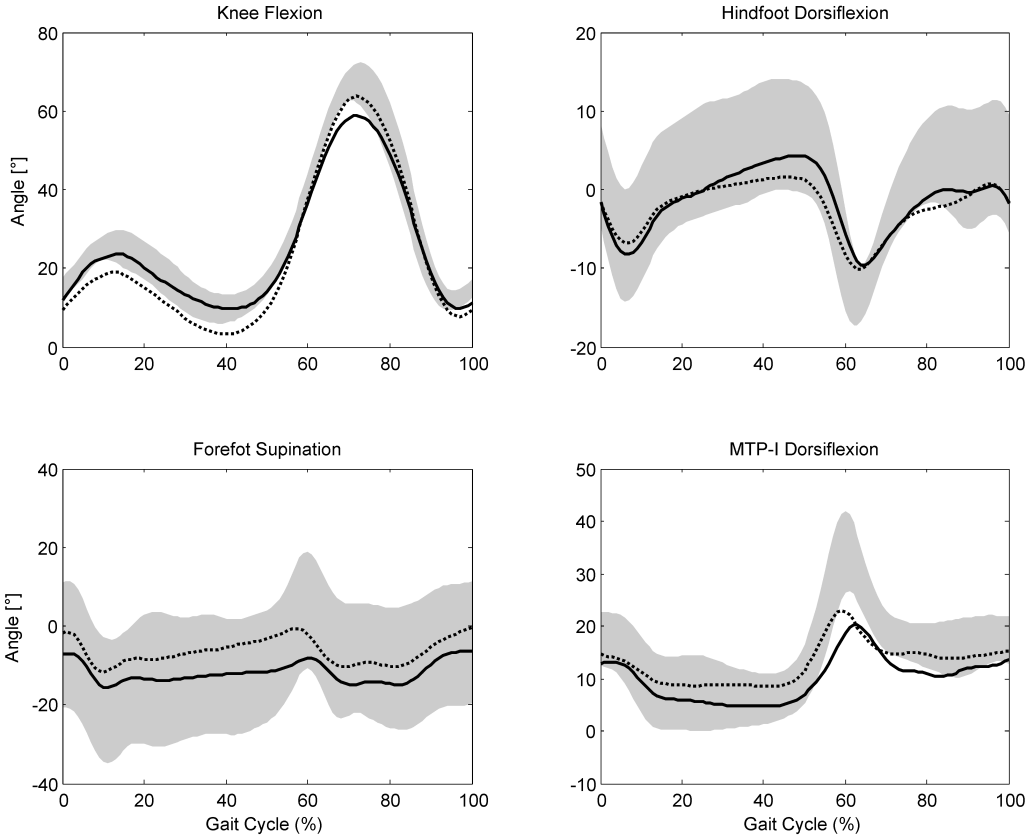

Figure 1: Mean knee flexion, hindfoot dorsiflexion, forefoot supination and metatarsophalangeal-I (MTP-I) dorsiflexion angles during one gait cycle (start and end at heel strike) for the control group (grey shaded area: mean ± standard deviation), patients with ankle OA (solid line), and patients after realignment surgery (dashed line).

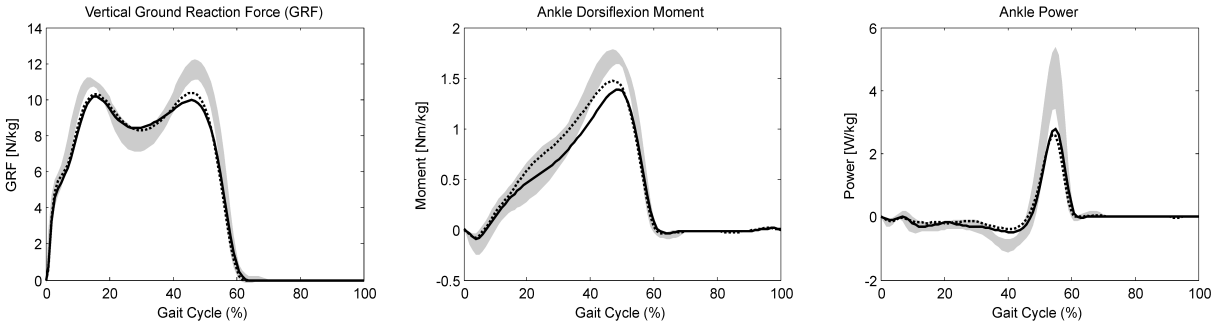

Figure 2: Mean vertical ground reaction force, ankle dorsiflexion moment, and ankle power during one gait cycle (start and end at heel strike) for the control group (grey shaded area: mean ± standard deviation), patients with ankle OA (solid line), and patients after realignment surgery (dashed line).

Table 7: Clinical results. Data are presented as a mean (range).

Parameter	Patients after realignment surgery	Patients with asymmetric ankle OA	Control group	*P* value
Dorsiflexion (°)	0.0	6.2	19.4	0.001†
	(-10 to 15)	(-10 to 15)	(15 to 25)	0.171; < 0.001; 0.005‡
Plantar Flexion (°)	44.4	51.9	63.1	0.027†
	(25 to 65)	(35 to 70)	(50 to 75)	0.485; 0.021; 0.210‡
Maximal calf circumference (cm)	37.9 (32.5 to 44.0)	37.7 (35.0 to 41.0)	37.9 (33.0 to 44.0)	0.983†
Side difference calf circumference (cm)	1.8 (0.5 to 5.0)	1.6 (0 to 3.0)	0.3 (0 to 1.0)	0.041† 0.952; 0.052; 0.095‡
SF-36	80.7	55.8	91.2	< 0.001†
	(36.0 to 95.1)	(33.9 to 71.9)	(84.7 to 96.5)	0.004; 0.283; < 0.001‡
Pain (VAS 0-10)	2.1	4.6	0	< 0.001†
	(0 to 3.5)	(2.0 to 7.0)		0.001; 0.009; < 0.001‡
AOFAS Hindfoot score	83.5	64.5	100	< 0.001†
	(72 to 100)	(26 to 76)		0.008; 0.021; < 0.001‡

†using ANOVA (analysis of variance) test

‡using Tukey HSD (honestly significant difference) post hoc test: patients after realignment surgery vs. patients with asymmetric ankle OA; patients after realignment surgery vs. control group; patients with asymmetric ankle OA vs. control group

4. Discussion

Supramalleolar osteotomies are increasingly recommended for treatment asymmetric varus [17,26,29,56] and valgus [17,20,56] ankle OA. In 2007 we reported mid-term results of realignment surgery in 35 consecutive patients with asymmetric posttraumatic ankle OA [18]. Realignment surgery was able to provide a significant pain relief and functional improvement as assessed using AOFAS hindfoot score and postponed the planned ankle fusion or total ankle replacement in 91% of the patient group [18]. However, to our knowledge, no clinical studies address gait outcomes in patients who underwent realignment surgery due to asymmetric ankle OA.

This study showed that the spatiotemporal variables of gait are substantially improved after realignment surgery versus the OA group. Similar findings with increased walking speed, higher cadence, and lower stride and step times have been observed in patients who underwent total ankle replacements [13,36,38]. With the higher walking speed one may expect a higher ROM, moments and power [57], yet this was not the case in the present study (Table 5, 6). On the contrary, the hindfoot dorsiflexion ROM was lower in the patients after realignment surgery. A possible explanation for this decreased hindfoot dorsiflexion ROM is that these patients also had the lowest passive ankle ROM (Table 7).

Many studies have used a simple one segment foot model to assess the biomechanical outcome of total ankle replacements or ankle arthrodesis [32-36,38]. In the present study, a more detailed foot model measured hindfoot, forefoot, and hallux kinematics. After surgery the sagittal ROM was not only reduced for the hindfoot, but also for the hallux. Both increased[58] and decreased[37] dynamic forefoot ROM have been observed in ankle arthrodeses. Our results for forefoot dorsiflexion indicate that the reduced passive ankle ROM was not compensated

for by increasing the motion of other joint in the foot. It has also been shown that an ankle ROM of 30° is sufficient for normal gait [59] and this ROM was achieved in most of the patients in the present study.

Despite the lower hindfoot dorsiflexion ROM, patients after realignment surgery had a comparable speed as the control subjects, which may partially be explained by the increased the flexion-extension ROM of the knee (Table 7). Further, the postoperative patient group had a higher hip flexion ROM, peak external hip extension moment, and peak external knee extension moment than the preoperative group though the differences were not significant. This could be an indication that the patients adapted a different gait pattern after surgery and support the lifting and clearing of the foot after toe-off by increasing the sagittal motion of the knee and hip. Alternatively, increased hip joint moments could be due to the higher walking speed in the postoperative group [57].

The peak external ankle dorsiflexion moment and peak ankle power were reduced compared to the healthy controls (Table 6). Brodsky et al.[13] showed an increase in the peak external dorsiflexion moment and peak ankle power at least two years after total ankle replacement, but other studies showed no change or decreases at least six months after total ankle replacement [33,34,38]. For ankle arthrodesis a slight increase in the peak ankle moment was found [33]. The second vertical peak of the ground reaction force (during push-off) was reduced in the present study for both patient groups (Table 6). One study found a similar result for total ankle replacements [38], but other results showed a higher peak ground reaction force [36].

In the present study, patients who underwent realignment surgery had significantly less pain as assessed using VAS and higher AOFAS hindfoot and SF-36 scores. This is comparable to clinical studies addressing outcomes following realignment surgery [17,18,26,27,60,61]. Takakura et al.[28,29,62] described substantial pain relief and reduced limitations in daily activities in their cohort. Cheng et al.[63] reported good or excellent results in 18 patients. Knupp et al.[17] established a novel classification of supramalleolar deformities and found significant postoperative pain relief and functional improvement in the AOFAS hindfoot score in 94 ankles. In our previous study significant pain relief and functional improvement were observed in 35 patients who underwent realignment surgery [18]. Furthermore, total ankle replacement or ankle fusion was postponed in 91% of all patients at the mean follow-up of 5 years [18].

Our study has some limitations. First, the number of patients who underwent realignment surgery in the present study is small, but the inclusion criteria (patients with isolated supramalleolar valgus deformity and supramalleolar osteotomy without additional osseous/ligamental procedures) were very strict to help avoid bias from confounding factors. Second, the patients who underwent realignment surgery were on average 10 years younger than the patients in the osteoarthritis group. It is possible that the higher walking speed in patients after realignment surgery was partially due to younger age. However, it has been shown that the self-selected walking speed is relatively constant for healthy individuals between 20 and 70 years of age [64]. Finally, functional outcome was partially assessed using

the AOFAS hindfoot score whose use has been recommended against [65]. The AOFAS score was in use before the policy statement was released so the data was not excluded. Alternative validated scores could be used in the future.

In conclusion, this study showed that the gait patterns of the patients who underwent realignment surgery were superior to those observed in patients with asymmetric ankle osteoarthritis. Although the ankle ROM decreased with respect to control subjects, patients in the realignment surgery group had less pain and higher quality of life than patients with osteoarthritis of the ankle joint. More than half of all patients with ankle joint osteoarthritis present with a malaligned hindfoot [2,4,66.] This makes supramalleolar osteotomies a promising treatment option in young and active individuals, since they may be able to postpone total ankle replacement and ankle arthrodesis [30] while restoring some physiologic gait parameters.

Acknowledgment:

The authors thank Beat Goepfert, University of Basel for technical support.

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Chapter 5

Effects of supramallolar osteotomies for ankle osteoarthritis on foot kinematics and lower leg muscle activation during walking

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Submitted to: Clinical Biomechanics

Background: Early stages of asymmetric ankle osteoarthritis can be treated by joint preserving supramalleolar osteotomies that surgically realign the ankle and unload the degenerated cartilage. While studies already showed pain relief and improvements in ankle scores effects on gait biomechanics are largely unknown. The purpose of this study was therefore to investigate the patients' gait pattern after supramalleolar osteotomies focusing on foot kinematics using the Oxford foot model and lower leg muscle activation.
Methods: An instrumented three-dimensional gait analysis with simultaneous electromyography (EMG) of five lower leg muscles (*Mm. gastrocnemius medialis* and *lateralis*, *soleus*, *peroneus longus,* and *tibialis anterior*) was performed with 12 patients with ankle osteoarthritis, seven short-term follow-up patients (12 – 18 months postoperatively), seven long-term follow-up patients (8 – 9 years postoperatively) and 15 healthy control subjects. The waveforms of the foot kinematics and EMGs were analyzed using principal component analysis.
Findings: Principal component (PC) scores that affected the sagittal range of motion of the hindfoot and hallux were reduced in all patient groups, while PC scores that affected the timing of the peaks in the sagittal forefoot motion were only reduced in preoperative and short-term follow-up patients. For the muscle activation, PC scores that changed the peak were lower for soleus and gastrocnemius medialis in preoperative and short-term follow-up patients.
Interpretation: Both postoperative patient groups showed similar changes in their gait pattern as patients with ankle osteoarthritis. These are probably related to a remaining reduction in ankle mobility. However this seems to affect the patients' well being less than a painful joint.

Keywords: ankle osteoarthritis; gait analysis; foot kinematics; electromyography; principal component analysis

1. Introduction

Ankle osteoarthritis (OA) mainly develops posttraumatically and often affects physically active, middle-aged people [1]. The patients' high expectations of maintaining a high level of physical activity and their commonly high life expectancy causes controversies regarding the optimal surgical treatment: The two main treatment options are ankle arthrodesis or ankle joint replacement. While ankle arthrodesis may facilitate a high physical activity level, it increases the chance of OA development in adjacent joints because of the reduced ankle mobility [2,3]. In contrast, ankle joint replacement retains the ankle mobility but complications might arise in revision surgery for implant failure because of the loss of bone stock [4].

In more than half of the patients with ankle OA a substantial malalignment of the hindfoot (most frequently varus) is present [1] leaving some regions of the articular cartilage still intact. One therefore often also speaks of asymmetric ankle OA. For such patients with asymmetric ankle OA, joint preserving realignment surgeries represent an alternative treatment option. With supramalleolar and hindfoot osteotomies (SMOT) the ankle is surgically realigned to reduce the load on the degenerated cartilage and to improve joint congruency. Several studies have shown that SMOT reduces pain and improves function measured by different functional ankle scores [5-7]. While about 40% of the patients with ankle OA reported no sports activity and about 40% a recreational sports activity of at least one hour per week, this changed following on average five years after SMOT to about 20% with no sports activity and about 70% with a recreational sports activity of at least one hour per week [8]. Further, the effect of SMOT on passive ankle range of motion is controversial. While one study reported an improvement [6], other studies found no changes or even a decrease in passive range of motion [9,10]. Although functional scores and passive range of motion are important, they do not reflect the actual biomechanics of the ankle during activities of daily living such as walking.

Gait analysis studies in patients with ankle OA have shown that patients walk slower and have a smaller range of motion in the ankle joint, hindfoot segment, and forefoot segment and have lower peak ground reaction forces, ankle dorsiflexion moment, and ankle power than healthy control subjects [11-13]. Contrary to SMOT, the effects of an ankle arthrodesis or total ankle replacement on the gait pattern have been investigated. These studies showed an increased walking speed with changes in the gait biomechanics after arthrodesis [14,15] and improvements in both walking speed and gait biomechanics after total ankle replacement [11,16]. In patients with ankle OA a prolonged activation of the calf muscles during walking has been observed [17,18]. After ankle arthrodesis soleus muscle activation during pre-swing when it is normally inactive was found [14]. To date, the functional outcome of SMOT in terms of biomechanical and neuromuscular gait data is largely unknown.

Principal component analysis (PCA) is a powerful tool for the analysis of gait waveform data [19-22]. Contrary to the conventional approach of reporting peak values, specific time points, or ranges of motion, PCA does not require any a-priori selection of the parameters of interest. However, the interpretation of the features that are derived from the PCA can be

challenging. Nevertheless, PCA has successfully been used to describe the postoperative gait patterns of knee OA patients after total knee replacement [20,22].

The aim of the present study was therefore to identify gait patterns that are typical for patients with ankle OA and after SMOT with the main focus on the foot kinematics and lower leg muscle activation. We hypothesized that after SMOT the walking speed, the peak ankle dorsiflexion moment and power were increased, and that the ankle range of motion remained decreased compared to healthy controls. Further, we expected that the features of the detailed foot kinematics and lower leg muscle activation patterns resolved by PCA better resemble those of healthy control subjects than those of patients with ankle OA.

2. Methods

2.1 Subjects

Three different groups of patients and a group of healthy controls participated in this study. The detailed subject characteristics are summarized in Table 1. The first group consisted of 12 patients with asymmetric (malalignment of the hindfoot) ankle OA that were eligible for realignment surgery. Exclusion criteria were OA in other joints, diabetes, neurological impairments, and neuromuscular diseases. Seven of these patients (prospective group) returned for a follow-up assessment between 12 and 18 months postoperatively. The long-term follow-up patient group consisted of seven different patients that had undergone realignment surgery eight to nine years before the gait analysis. Finally, data of a group of 15 healthy control subjects with a comparable age as the patients was collected as reference gait data. Exclusion criteria for the control group were pain in the lower extremities, repetitive ankle sprains, known OA in any joint of the lower extremity, diabetes, neurological impairments, and neuromuscular diseases. As previously reported, we defined the dominant leg of the controls as the preferred hopping leg [13]. The local ethics committee approved the study and all subjects signed informed consent prior to participation.

2.2. Clinical evaluation

The level of pain in the ankle joint was determined using a visual analogue scale ranging from 0 (no pain) to 10 (maximal pain) [23]. The clinical functional status was assessed with the American Orthopaedic Foot and Ankle Society (AOFAS) ankle and hindfoot score [24] (minimum score: 0 points; maximum score: 100 points). The healthy control subjects were only included if their pain score was 0 and their AOFAS score was 100. All subjects were further asked to complete the short-form-36 (SF-36) health survey questionnaire (version 2; Medical Outcomes Trust, Waltham, USA; minimum score: 0 points, maximum score: 100 points) providing a measure of general health and quality of life. The passive ankle range of motion in plantar- and dorsiflexion was measured using a goniometer.

Table 1: Subject characteristics

	Controls (*n* = 15)	**Ankle OA (*n* = 12)**	**Short-term follow-up (pre)* (*n* = 7)**	**Short-term follow-up (post) (*n* = 7)**	**Long-term follow-up (*n* = 7)**	***P* Value ANOVA**		
	Mean (95% CI)	**Mean (95% CI)**	**Mean (95% CI)**	**Mean (95% CI)**	**Mean (95% CI)**	**Model 1**	**Model 2**	**Model 3**
Age (yr) at time of gait analysis	48.5 (42.7, 54.4)	55.2 (47.9, 62.4)	49.9 (40.1, 59.7)	51.1 (41.4, 60.9)	46.3 (41.0, 51.7)	0.860	0.132	0.611
Height (m)	1.73 (1.66, 1.79)	1.75 (1.70, 1.81)	1.75 (1.68, 1.82)	1.75 (1.68, 1.82)	1.75 (1.62, 1.88)	0.867	0.823	0.997
Body mass (kg)	74.6 (69.2, 80.1)	88.0 (78.6, 97.4)	89.9 (79.3, 100.6)	89.4 (76.7, 102.2)	87.1 (61.6, 112.6)	0.005 0.004; 0.009; 0.776†	0.086	0.956
BMI (kg/m^2)	25.0 (23.6, 26.4)	28.5 (26.4, 30.6)	29.3 (26.8, 31.9)	29.3 (25.8, 32.8)	27.8 (23.6, 32.1)	0.002 0.002; 0.005; 0.957†	0.026 0.005; 0.074; 0.735†	0.705

*: This is a subgroup of the patients with ankle OA and includes those patients that returned for a follow-up measurement after surgery
ANOVA Model 1 included the following groups: controls, short-term follow-up (pre), and short-term follow-up (post)
ANOVA Model 2 included the following groups: controls, ankle OA, and long-term follow-up
ANOVA Model 3 included the following groups: short-term follow-up (pre), short-term follow-up (post), and long-term follow-up
†: Post-hoc tests were performed using a paired t test for the prospective comparison and an independent sample Student's t test for the other comparisons.

2.3 Gait analysis

A gait analysis system consisting of a six camera motion capture system (Vicon MX, Oxford, UK; sampling rate 120 Hz) and two force plates (Kistler, Winterthur, Switzerland; sampling rate 2400 Hz) was used to assess the subjects' gait patterns. Reflective markers were placed on anatomical landmarks according to the conventional gait model [25] and the Oxford foot model [26]. All subjects walked barefoot at their self-selected speed along a 10 m laboratory walkway until six trials were recorded where the subjects hit the force plates centrally. The gait data was processed using the Vicon Nexus software. Spatiotemporal variables, ankle range of motion, peak ankle dorsiflexion moment, peak ankle power, and the detailed temporal kinematics of the foot segments measured with the Oxford foot model were further analyzed. The detailed foot kinematic data included hindfoot (hindfoot to tibia) dorsiflexion, hindfoot inversion, forefoot (forefoot to hindfoot) dorsiflexion, forefoot supination, and hallux (hallux to forefoot) dorsiflexion angle. All these waveforms were time-normalized to one gait cycle, starting and ending with heel strike. Further, a symmetry index for step length and stance phase was calculated by dividing the value of the affected side by the value of the non-affected side (greater deviations from 1 indicate greater asymmetry).

2.4 EMG recordings

Surface electromyographic data (EMG) of *Mm. gastrocnemius medialis*, and *lateralis*, *soleus*, *peroneus longus*, and *tibialis anterior* was recorded bilaterally with a sampling rate of 2400 Hz (recording system: Biovision, Wehrheim, Germany. Bandwidth 10–700 Hz, gain range 1000–5000). Round, bipolar Ag/AgCl electrodes (Noraxon U.S.A. Inc., Scottsdale, AZ, USA: 10 mm diameter, 22 mm interelectrode distance) were placed according to the guidelines of the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM) project [27] after shaving and cleaning the skin with alcohol. The ground electrodes were placed bilaterally on the tibial tubercles. To minimize movement artifacts, amplifiers and cables were taped to the skin and held in place by an elastic net bandage around the calf.

The EMG signals were processed by rectifying the signal and applying a low-pass filter (4^{th} order, zero-lag Butterworth filter, 6 Hz cut off frequency) [20]. Each EMG envelope was then time-normalized to one gait cycle and amplitude-normalized to its mean amplitude [28].

2.5 Principal component analysis

PCA is a statistical tool for data reduction and has often been used in the analysis of gait waveform data [13,19-22]. It is an orthogonal transformation that transforms the correlated original data into uncorrelated orthogonal principal components (PC). The resulting PC vectors (features of the input data), and PC scores (individual loadings of the PC vectors) were further analyzed. To achieve data reduction, only the first few components that explained combined 70% of the total variance were retained [29].

The waveform data of the foot kinematics and EMG envelopes of the patients' affected leg and the controls' non-dominant leg were used as input. For each of these 10 sets of waveform

data (5 kinematic waveforms, 5 EMG waveforms) an input matrix X (101 time points x 246 trials (6 x 41 subjects)) was formed. The kinematic waveforms showed large offsets between measurements, i.e. parallel shifts of the curves [13]. Based on our experience this offset will mainly be captured by the first PC and could explain up to 90% of the variance. Because we were more interested in the shape of the waveform than in its magnitude, we minimized this offset by subtracting the mean angle over time for each trial from each waveform data before calculating the PCA.

2.6 Statistics

All statistical tests were carried out in MATLAB (Version 7.10, Mathworks, Natick, MA, USA). For each subject the gait parameters and PC scores were averaged over the six walking trials. Differences between the patient groups and the controls were detected using three different ANOVA models with a significance level of $\alpha = 0.05$. The first model (model 1) included the controls, preoperative short-term follow-up patients, and the postoperative short-term follow-up patients and assessed the effect of the surgery on the patients' gait pattern. Model 2 was used to assess the long-term effect of the surgery and included the controls, patients with ankle OA, and the long-term follow-up patients. Finally, model 3 used the preoperative short-term follow-up patients, the postoperative short-term follow-up patients, and the long-term follow-up patients. Student's independent samples t tests and paired t tests (prospective data) were used for the post hoc analysis of the differences between the groups. The significance level was adjusted to multiple comparisons and set at $\alpha = 0.017$.

3. Results

3.1 Clinical scores and gait parameters

After surgery, patients reported less pain than patients with ankle OA both in the short-term follow-up group ($p = 0.001$) and long-term follow-up group ($p = 0.005$). The AOFAS scores were increased by 15 points in the short-term follow-up group ($p = 0.047$), and by 20 points in the long-term follow-up group ($p = 0.006$). The SF-36 score was significantly reduced compared to the controls in the patients with ankle OA ($p < 0.001$), and short-term follow-up patients both preoperatively ($p < 0.001$), and postoperatively ($p = 0.011$), but not in the long-term follow-up patients. Further, all patient groups had significantly reduced passive ankle plantar- and dorsiflexion ranges of motion ($p < 0.001$) with a trend towards an additional postoperative reduction in the prospective patients ($p = 0.038$) (Figure 1).

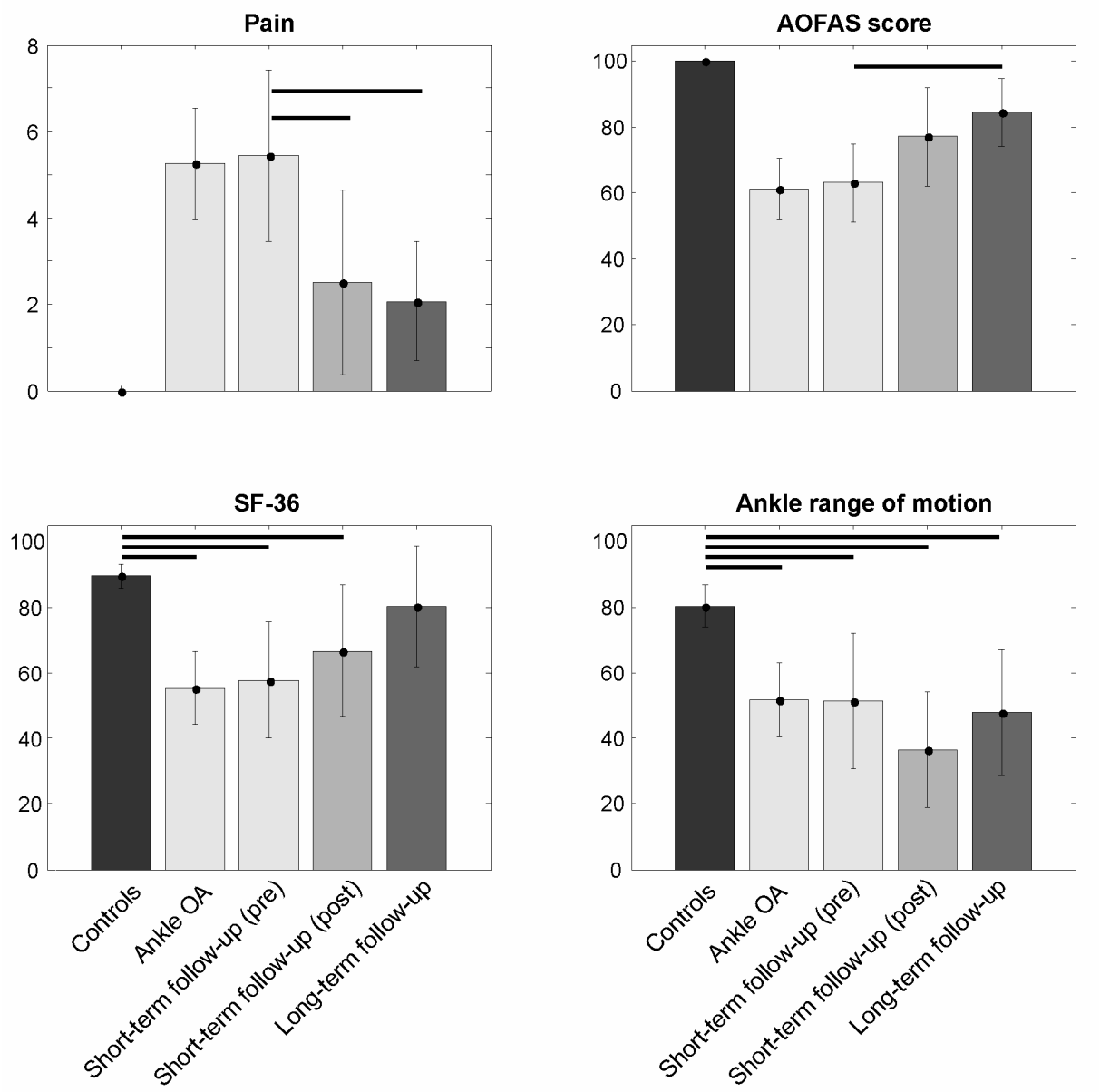

Figure 1: Clinical scores and passive ankle range of motion presented as means (bar) and 95% confidence interval of the mean (error bar). Vertical lines indicate significant differences between the groups according to the three ANOVA models.

The self-selected walking speed was significantly reduced compared to the controls by about 15% in the patients with ankle OA, the preoperative short-term follow-up, and the postoperative short-term follow-up patients. The step length and the length of the stance phase showed no significant differences, but there was a greater asymmetry between the affected and non-affected side than in the controls (Table 2). Further it was seen that on average the ankle range of motion was reduced in all patient groups by at least 25%, the peak dorsiflexion moment by at least 15% and the ankle power by at least 40% compared to the control subjects (Table 2).

3.2 Effects on the gait pattern

The mean angles of the foot segments of the different groups are depicted in Figure 2. The visual inspection of the mean temporal waveforms from the muscle activation of the lower leg muscles showed only minor differences between the healthy controls and the different patient groups. The main difference was the lower peak activity especially of *Mm. gastrocnemius medialis* and *soleus* (Figure 3). These waveforms were analyzed by PCA and Table 3 summarizes the features of the foot kinematics and EMG envelopes that showed significant differences between the groups (Table 3).

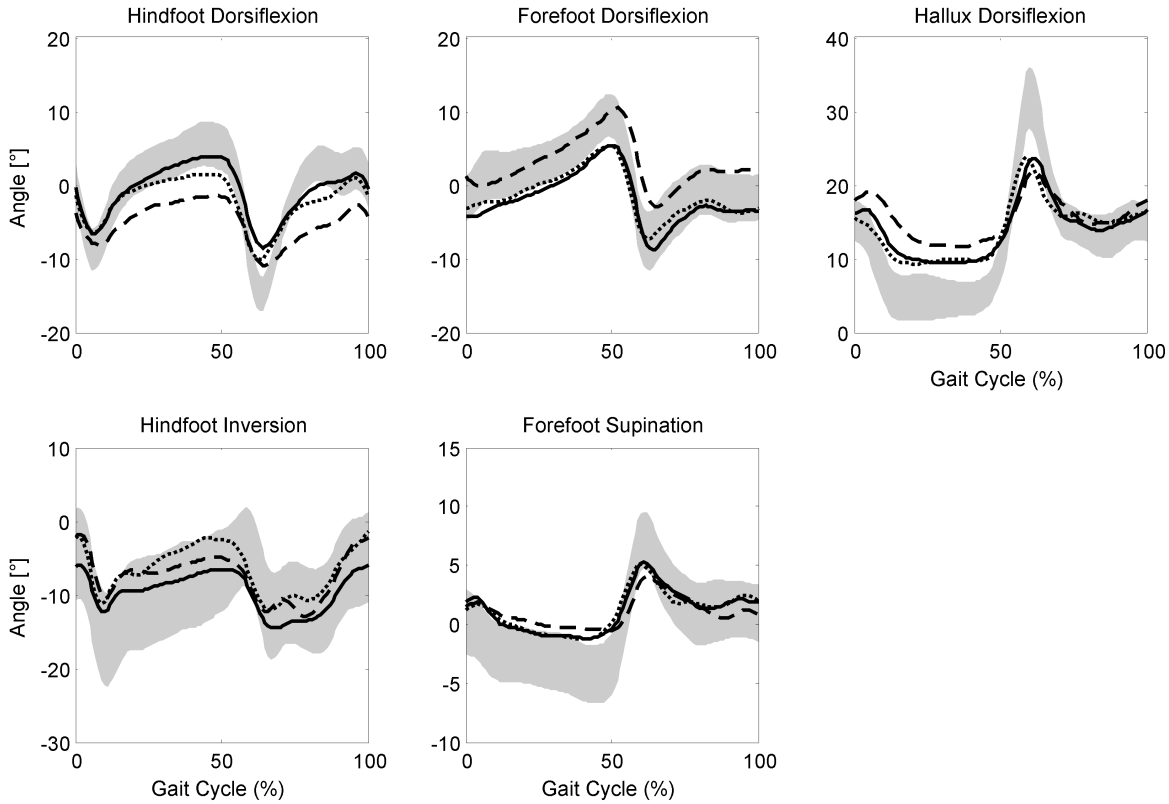

Figure 2: Foot kinematic angles of the healthy controls (grey shaded area: mean and 95% confidence interval of the mean), the patients with ankle OA (solid line), short-term follow-up patients (dashed line), and long-term-follow-up patients (dotted line). All angles start and end at heel strike.

3.2.1 Short-term effect on the gait pattern (prospective data)

In the prospective foot kinematics data, two PCs showed significantly worse results after surgery: postoperative short-term follow-up patients had an even lower hindfoot plantarflexion range of motion during push-off (PC1) and lower amplitude in hindfoot inversion (PC2) than preoperatively. The following features of the foot kinematics showed significant changes at both measurement time points compared to controls: hindfoot dorsiflexion amplitude (PC2), timing of the hindfoot inversion (PC1), timing of the forefoot dorsiflexion (PC2), amplitude of the forefoot supination (PC1), and amplitude of the hallux dorsiflexion (PC1). For the muscle activation, both patient groups had significantly lower peak activation of *M. gastrocnemius medialis* (PC1). The peak activity of *M. soleus* (PC1)

and the amplitude of the *M. tibialis anterior* activation (PC1) were only significantly lower in the preoperative patients. The peak activation of *M. tibialis anterior* at the beginning of the swing phase (PC3) was significantly lower postoperatively, compared to preoperatively (Table 3).

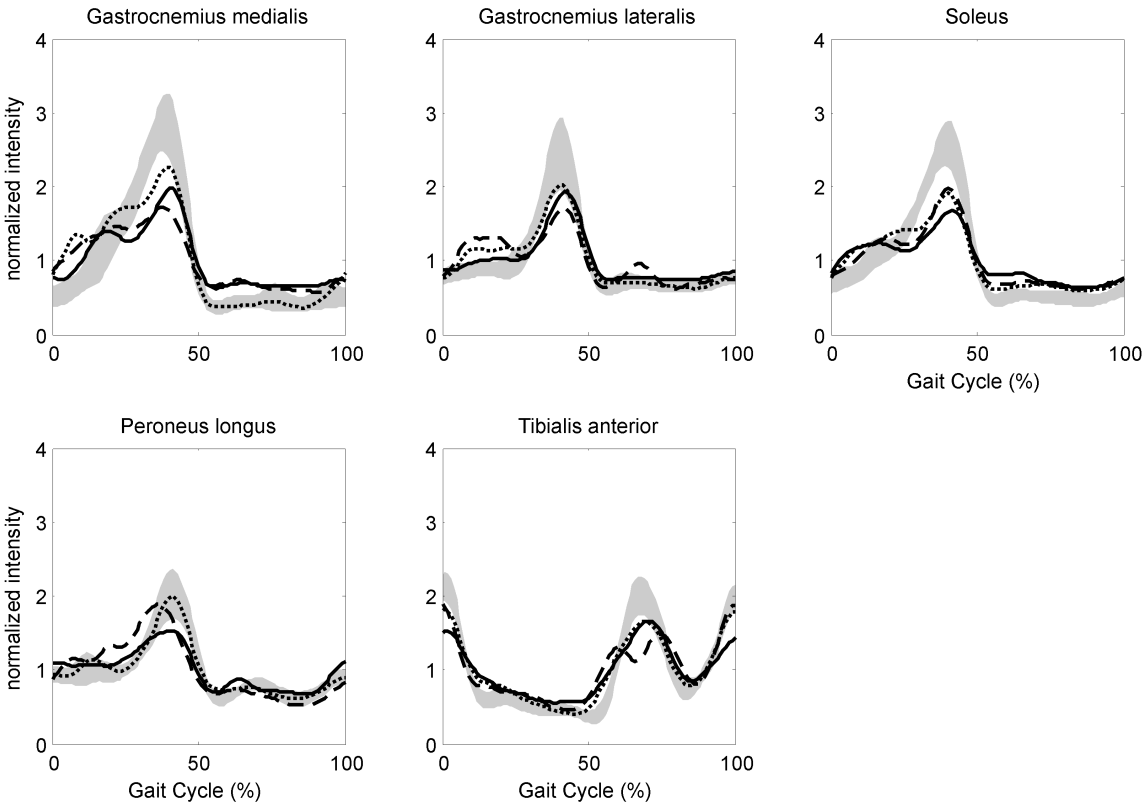

Figure 3: Mean of the normalized EMG envelopes of *Mm. gastrocnemius medialis* and *lateralis*, *soleus*, *peroneus longus*, and *tibialis anterior* for the healthy controls (grey shaded area: mean and 95% confidence interval of the mean), the patients with ankle OA (solid line), postoperative short-term follow-up patients (dashed line), and long-term-follow-up patients (dotted line). All EMG envelopes start and end at heel strike.

3.2.2 Long-term effect on the gait pattern

Both patients with ankle OA and long-term follow-up patients had significantly different PC scores for the hindfoot plantarflexion range of motion during push-off (PC1), timing of the hindfoot inversion (PC1), amplitude of the forefoot supination (PC1), and amplitude of the hallux dorsiflexion (PC1). The PCA of the EMG envelopes only showed significant differences between the controls and the patients with ankle OA with a lower peak for *Mm. gastrocnemius medialis* (PC1) and *soleus* (PC1), and a lower amplitude for *M. tibialis anterior* (PC1) (Table 3).

Table 2: Gait parameters

	Controls (*n* = 15)	**Ankle OA (*n* = 12)**	**Short-term follow-up (pre)* (*n* = 7)**	**Short-term follow-up (post) (*n* = 7)**	**Long-term follow-up (*n* = 7)**	***P* Value ANOVA**		
	Mean (95% CI)	**Mean (95% CI)**	**Mean (95% CI)**	**Mean (95% CI)**	**Mean (95% CI)**	**Model 1**	**Model 2**	**Model 3**
Walking speed (m/s)	1.26 (1.20, 1.32)	1.07 (0.95, 1.18)	1.08 (0.88, 1.28)	1.03 (0.79, 1.27)	1.19 (1.11, 1.26)	0.017 0.016; 0.007; 0.232†	0.004 0.002; 0.124; 0.123†	0.359
Cadence (steps/min)	112.49 (107.61, 117.17)	103.47 (99.18, 107.76)	106.02 (99.43, 112.61)	104.49 (96.24, 112.74)	107.66 (102.80, 112.53)	0.087	0.015 0.007; 0.200; 0.177†	0.720
Step length (m)	0.68 (0.64, 0.71)	0.63 (0.58, 0.69)	0.62 (0.52, 0.72)	0.62 (0.52, 0.73)	0.67 (0.63, 0.71)	0.288	0.262	0.541
% of stance phase	60.99 (60.22, 61.76)	60.77 (59.82, 61.72)	61.35 (60.16, 62.53)	61.24 (59.86, 62.63)	59.98 (58.57, 61.38)	0.835	0.323	0.169
Symmetry index step length	1.00 (0.98, 1.02)	1.07 (1.00, 1.14)	1.08 (0.98, 1.17)	1.14 (1.02, 1.27)	1.03 (0.99, 1.06)	0.004 0.017; 0.001; 0.089†	0.046 0.024; 0.126; 0.299†	0.119
Symmetry index stance phase	1.00 (0.99, 1.01)	0.96 (0.94, 0.98)	0.97 (0.95, 0.99)	0.96 (0.92, 1.00)	0.96 (0.94, 0.98)	0.006 0.004; 0.005; 0.403†	<0.001 0.001; <0.001;0.957†	0.692
Ankle range of motion (°)	31.05 (28.74, 33.36)	23.67 (18.71, 28.63)	23.37 (14.01, 32.73)	21.44 (12.61, 30.27)	19.54 (15.72, 23.37)	0.013 0.019; 0.003; 0.440†	<0.001 0.004; <0.001; 0.215†	0.700
Peak ankle dorsiflexion moment (Nm/kg)	1.74 (1.68, 1.80)	1.36 (1.14, 1.59)	1.38 (1.08, 1.68)	1.06 (0.84, 1.28)	1.49 (1.32, 1.66)	<0.001 0.001; <0.001; 0.002†	0.001 0.001; 0.001; 0.397†	0.016 0.456; 0.003; 0.002†
Peak ankle power (W/kg)	4.74 (4.38, 5.10)	2.82 (1.82, 3.83)	2.99 (1.31, 4.66)	2.09 (1.04, 3.15)	2.92 (2.17, 3.67)	<0.001 0.003; <0.001; 0.047†	<0.001 <0.001; <0.001; 0.885†	0.386

*: This is a subgroup of the patients with ankle OA and includes those patients that returned for a follow-up measurement after surgery
ANOVA Model 1 included the following groups: controls, short-term follow-up (pre), and short-term follow-up (post)
ANOVA Model 2 included the following groups: controls, ankle OA, and long-term follow-up
ANOVA Model 3 included the following groups: short-term follow-up (pre), short-term follow-up (post), and long-term follow-up
†: Post-hoc tests were performed using a paired t test for the prospective comparison and an independent sample Student's t test for the other comparisons

Table 3: Mean principal component (PC) scores with the corresponding features of the foot kinematics and EMG waveforms

Features (PC, explained variance (%))	Controls (*n* = 15)	Ankle OA (*n* = 12)	Short-term follow-up (pre)* (*n* = 7)	Short-term follow-up (post) (*n* = 7)	Long-term follow-up (*n* = 7)	*P* Value ANOVA		
	Mean (95% CI)	Mean (95% CI)	Mean (95% CI)	Mean (95% CI)	Mean (95% CI)	Model 1	Model 2	Model 3
Range of motion **hindfoot plantarflexion** during push-off (PC1, 41.4%)	13.14 (5.46, 20.82)	-5.93 (-16.41, 4.54)	1.17 (-12.45, 14.79)	-12.46 (-25.07, 0.16)	-5.53 (-12.54, 1.49)	0.002 0.079; 0.001; 0.009†	0.002 0.003; 0.004; 0.952†	0.149
Amplitude **hindfoot dorsiflexion** (PC2, 26.6%)	6.90 (0.40, 13.40)	0.48 (-9.54, 10.49)	-9.21 (-14.25, -4.18)	-7.51 (-15.98, -0.96)	-8.09 (-19.10, 2.93)	0.001 0.003; 0.010; 0.587†	0.061	0.940
Timing peak **hindfoot plantarflexion** at push-off (PC3, 15.6%)	-4.97 (-10.55, 0.62)	3.05 (-2.68, 8.79)	5.41 (-3.32, 14.15)	6.75 (-1.98, 15.48)	-1.34 (-5.30, 2.62)	0.019 0.033; 0.018; 0.675†	0.080	0.163
Timing peak **hindfoot inversion** push-off, and magnitude inversion in stance (PC1, 39.8%)	-12.95 (-19.11, -6.79)	7.17 (-4.98, 19.32)	4.62 (-12.16, 21.44)	6.88 (-7.84, 21.59)	8.59 (-8.18, 25.35)	0.006 0.011; 0.003; 0.541†	0.003 0.002; 0.002; 0.876†	0.914†
Amplitude **hindfoot inversion** (PC2, 16.9%)	6.00 (1.10, 10.91)	-2.81 (-10.72, 5.11)	2.90 (-4.40, 10.19)	-8.20 (-14.45, -1.94)	0.15 (-10.75, 11.05)	0.003 0.438; 0.001; 0.002†	0.118	0.085
Timing **forefoot dorsiflexion** and plantarflexion peaks during push-off (PC2, 19.7%)	-4.75 (-8.75, -0.75)	2.35 (-2.89, 7.59)	3.87 (-4.99, 12.72)	9.88 (-2.34, 22.10)	-3.73 (-12.34, 4.88)	0.006 0.029; 0.003; 0.115†	0.076	0.090
Amplitude **forefoot supination** (PC1, 41.3%)	9.77 (5.98, 13.55)	-3.55 (-9.89, 2.79)	-1.83 (-9.91, 6.24)	-10.15 (-16.73, -3.57)	-4.70 (-11.88, 2.48)	<0.001 0.003; <0.001; 0.025†	<0.001 <0.001; <0.001; 0.797†	0.164
Amplitude **hallux dorsiflexion** (PC1, 43.6%)	24.73 (16.06, 33.41)	-10.91 (-23.68, 1.85)	-6.51 (-24.61, 11.59)	-24.21 (-38.43, -9.99)	-10.08 (-25.36, 5.21)	<0.001 0.001; <0.001; 0.007†	<0.001 <0.001; <0.001; 0.927†	0.156
Peak activity ***M. gastrocnemius medialis*** (PC1, 44.8%)	2.20 (0.38, 4.03)	-1.26 (-3.32, 0.79)	-1.75 (-4.26, 0.76)	-2.20 (-4.20, -0.19)	-0.42 (-3.68, 2.84)	0.003 0.012; 0.004; 0.526†	0.030 0.011; 0.104; 0.603†	0.495
Peak activity ***M. soleus*** (PC1, 53.0%)	2.02 (0.72, 3.33)	-1.69 (-2.99, -0.39)	-1.62 (-3.90, 0.67)	-0.65 (-2.94, 1.64)	-0.78 (-4.17, 2.62)	0.005 0.003; 0.024; 0.467†	0.002 <0.001; 0.042; 0.489†	0.800
Amplitude activity ***M. tibialis anterior*** (PC1, 33.6%)	1.22 (0.22, 2.21)	-0.89 (-2.25, 0.47)	-1.38 (-3.69, 0.92)	-0.83 (-3.51, 1.84)	-0.25 (-1.90, 1.40)	0.032 0.011; 0.054; 0.482†	0.024 0.010; 0.089; 0.512†	0.688
Timing peak activity ***M. tibialis anterior*** at beginning of swing phase (PC2, 18.4%)	-0.75 (-1.51, 0.01)	0.61 (-0.69, 1.91)	1.04 (-0.81, 2.89)	0.73 (-0.54, 1.99)	-0.17 (-1.54, 1.19)	0.028 0.023; 0.029; 0.551†	0.123	0.375
Peak activity ***M. tibialis anterior*** at beginning of swing phase (PC3, 13.5%)	0.13 (-0.46, 0.72)	0.50 (-0.51, 1.50)	0.88 (-0.49, 2.24)	-1.08 (-2.29, 0.14)	-0.04 (-0.93, 0.85)	0.020 0.189; 0.032; 0.008†	0.621	0.033 0.193; 0.117; 0.008†

*: This is a subgroup of the patients with ankle OA and includes those patients that returned for a follow-up measurement after surgery
ANOVA Model 1 included the following groups: controls, short-term follow-up (pre), and short-term follow-up (post)
ANOVA Model 2 included the following groups: controls, ankle OA, and long-term follow-up
ANOVA Model 3 included the following groups: short-term follow-up (pre), short-term follow-up (post), and long-term follow-up
†: Post-hoc tests were performed using a paired t test for the prospective comparison and an independent sample Student's t test for the other comparisons.

4. Discussion

This study describes the gait patterns of patients with asymmetric ankle OA and of two patient groups after treatment with SMOT in comparison to those of healthy controls. Our results on the clinical outcome of SMOT of reduced pain and increased AOFAS ankle scores are in agreement with previous studies [5-7,9,10]. However, the hypotheses that the walking speed would be greater and gait patterns would improve postoperatively were only partly supported by our results. The hypothesis that the peak ankle dorsiflexion moment and ankle power would be greater postoperatively had to be rejected.

The long-term follow-up patients walked with similar walking speed, cadence and step length symmetry as the healthy controls whereas these parameters differed in short-term follow-up patients. A possible reason for these different outcomes in the two postoperative groups is that the follow-up time of around one year was too short and that these patients were still adjusting to the changed alignment and biomechanics of the ankle joint. Further, the long-term follow-up patients were on average five to ten years younger and hence much younger at the time of surgery than the short-term follow-up group. It is possible that the long-term follow-up patients had a higher recovery potential because of their younger age. Another important aspect of these results is that the positive effects of SMOT on the spatiotemporal parameters and the clinical scores (pain, AOFAS, SF-36) were sustained for at least eight years.

The changes in the foot kinematics were similar for both postoperative groups with reduced amplitudes and ranges of motion in the different foot segments. According to Perry [30], normal gait requires a minimal ankle range of motion of 30°. While most of the follow-up patients achieved this, three patients had a passive range of motion of 20° or less. On average, the lowest passive ankle ranges of motion were found in the short-term follow-up patients that also had the lowest PC scores in PCs that affected the hindfoot and hallux range of motion. Hence, it is likely that the observed changes are related to the reduced passive range of motion in the ankle joint. For a successful realignment of the ankle joint, a restriction of the joint mobility is often necessary to prevent recurrence of the hindfoot malalignment. Contrary to SMOT, ankle arthrodesis eliminates the motion in the ankle joint and often the subtalar and forefoot joints take over some of the plantar- and dorsiflexion movement of the foot. As Wu et al., [14] showed in a gait analysis study with patients after ankle arthrodesis, the shape of the hindfoot to tibia dorsiflexion angle is significantly altered. In a normal ankle, the hindfoot plantarflexes after heel strike and then gradually dorsiflexes until about 50% of the gait cycle. Then during push-off the hindfoot plantarflexes and moves towards dorsiflexion during the swing phase (Figure 2). However, after ankle arthrodesis, the hindfoot dorsiflexion angle remains constant throughout most of the stance phase [14]. In our study, using a comparable foot model, the hindfoot dorsiflexion waveform in patients with ankle OA and in patients after SMOT was similar to that in the control subjects (Figure 2). However, the PC2 scores indicated that both follow-up groups had a lower amplitude and thus also a less steep slope of the curve (Table 2). Differences in the forefoot dorsiflexion angle were only found for the timing but not for the range of motion. This result indicates that none of the patient groups

compensated for the reduced hindfoot range of motion with an increased forefoot range of motion (Table 2).

Contrary to our initial hypothesis, the peak kinetic values showed no improvement following SMOT. These values are influenced by the walking speed [31], range of motion, and the push-off force. Although that the long-term follow-up patients walked faster than the patients with ankle OA, their average peak ankle moment was only slightly higher. Further the ankle power in all patient groups was on average at least 40% lower than the value of the controls. While pain and muscle weakness have been suggested to lead to lower ground reaction forces during push-off in patients with ankle OA [13], the results for the follow-up patients indicate that the reduced ankle range of motion might also limit the force production during push-off.

The temporal muscle activation patterns differed only slightly between the groups. The main difference in muscle activation between patients and control subjects was the peak activation of *Mm. gastrocnemius medialis* and *soleus*. This result is not surprising since it is known that patients with ankle OA suffer from lower leg muscle atrophy, especially of *M. soleus* [32,33]. Even after total ankle joint replacement, this atrophy only partly reverses as indicated by reduced calf circumference in the affected compared to the unaffected side [32]. This might explain the reduced peak activation in the investigated calf muscles. However, it is important to note that we normalized the EMG envelopes to their mean amplitude, which reduces the differences in absolute values. Hence, our results indicate that the peak activation in patients differed less from the baseline activation or that their muscle was activated longer. Indeed, longer periods of muscle activation have previously been observed in patients with ankle OA [17,18].

The number of patients in the two postoperative groups was rather small. However, ankle OA has a much lower prevalence than OA of the knee or hip [34] and only patients with asymmetric ankle OA are eligible for SMOT. Therefore, only a small number of patients qualified for inclusion in our study, which limits the generalization of the results. Nevertheless, the data presented in this study represent the first data on gait biomechanics after SMOT.

In conclusion, the results showed that patients with ankle OA benefit from SMOT, mainly because of the associated pain relief. The gait biomechanics of the different foot segments presumably are strongly influenced by the passive ankle range of motion, which was lower in short-term follow-up patients compared to control subjects and was similar between patients with ankle OA and long-term follow-up patients. SMOT represents an early surgical treatment option for ankle OA that does not preclude further treatment with total ankle replacement or ankle arthrodesis. Especially, for younger patients postponing these limiting procedures is very important and our results are promising because pain relief and improved quality of life were maintained for at least eight years postoperatively despite of altered gait patterns.

Acknowledgment:

The authors thank the Swiss National Science Foundation (SNF No. 3200BO-120631) for financial support of the study and Beat Goepfert for technical support and help with the data collection.

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Chapter 6

General discussion and outlook

1. Discussion

The first aim of this thesis was to characterize the gait patterns of patients with early- to mid-stage, asymmetric ankle osteoarthritis (OA). These patients showed similar gait adaptations as patients with end-stage ankle OA as it was described in Chapters 2 and 3. Contrary to a previous study on patients with ankle OA that found shifts towards lower electromyography (EMG) frequencies in all lower leg muscles (*Mm. gastrocnemius medialis, soleus, peroneus longus, and tibialis anterior*) [1,2], this was not the case for the patients that participated in the measurements for this thesis (Chapter 3). One possible explanation could be that changes in the muscle activation, which lead to lower EMG frequencies, were not yet present in the patients with early- to mid-stage asymmetric ankle OA. However, there are also other factors (e.g. electrode position, amount of fat tissue) that influence the frequency content of the EMG signal [3] and that could lead to the observed changes in the power spectra. Secondly, this thesis aimed to assess the biomechanical and neuromuscular rehabilitation potential after treatment of the ankle OA with supramalleolar osteotomies. Compared to healthy subjects, the spatiotemporal parameters showed no differences in a long-term follow-up group (Chapter 4), while the range of motion, especially of the hindfoot, remained reduced in both a short-term, and long-term follow-up group (Chapter 4 and 5). This was likely related to the reduced mobility of the ankle joint. However, despite the remaining changes in the patients' gait pattern after supramalleolar osteotomies, patients reported less pain and had a higher quality of live. Many gait analysis studies on patients with ankle OA and on the treatment effects (ankle arthrodesis or total ankle replacement) used the conventional gait model [4] which models the foot as a single segment [5-7]. For this thesis, we used the Oxford foot model [8] that consists of three foot segments - hindfoot, forefoot, and hallux - and therefore provides more detailed information on the movement of the foot during walking. Contrary to some studies on ankle arthrodesis [9] that showed an increased motion of the forefoot, the results of this thesis indicate that the patients had enough flexibility of the ankle to allow walking without increasing the range of motion in the forefoot (Chapter 4 and 5).

In most of the patients, ankle OA has a posttraumatic etiology [10,11]. Therefore, it often affects younger and physically active patients [12,13]. Both common treatment options – ankle arthrodesis and total ankle replacement – have some concerns regarding the long-term outcome. For ankle arthrodesis, it has been shown that almost all patients develop OA in the neighbouring joints of the foot within 20 years [14,15]. For total ankle replacements, a survival rate of only about 80% after 10 years has been shown [16]. This is less than what is seen for knee and hip replacements [17,18]. Furthermore, revision surgeries in case of implant failure might be difficult due to the limited bone stock in the ankle joint [19]. Therefore, alternative treatment options are important. Joint preserving, realignment surgeries (supramalleolar osteotomies) are such an option for patients with asymmetric ankle OA and only partially degenerated articular cartilage. The patients' gait biomechanics that were observed in this thesis before and after realignment surgery showed that the gait patterns do not return to those of healthy subjects but that some parameters (e.g. walking speed) improve compared to the preoperative situation. The observed difference between the short- and long-term outcomes (Chapter 5) could indicate that the rehabilitation process was still ongoing one year

postoperatively and that an additional measurement two years after surgery could be helpful to assess the rehabilitation potential. The long-term outcome showed that a successful realignment surgery was able to postpone a total ankle replacement or an ankle arthrodesis by at least about eight years with gait patterns that are comparable to those after total ankle replacements [5-7,20] or ankle arthrodesis [7,9,20,21] (Chapter 4 and 5).

1.1 Comments on the measurement methods

There were only few patients included in the measurements for this thesis. Ankle OA is less frequent than hip or knee OA [22] and not all patients with ankle OA were eligible for realignment surgery and could be included in the study. Despite the low number of patients significant differences between healthy subjects and patients with asymmetric ankle OA, as well as effects of the treatment were found. However, it remains unclear whether these results can be generalized on the whole population of patients with ankle OA and whether all patients show similar changes in their gait pattern after realignment surgery.

1.1.1 Oxford foot model

There are many foot models around at the moment. For this thesis, we choose the Oxford foot model [8]. This model has been validated, it was available together with our measurement system, and it has become increasingly popular. Since the Oxford foot model has been developed that it can also measure feet with deformities, it does not reference the joint angles to a static calibration position [8]. During our measurements we sometimes observed large offsets (parallel shifts) of the different foot kinematic waveforms between subjects or between the two feet of a subject. This was not relevant for the analysis of the range of motion (Chapter 2 and 4). However, the analysis of these waveforms with the principal component analysis (PCA) showed that the first principal component (PC) captured this offset (Chapter 2 and 5). The larger the offset or the variation between the different waveforms, the more of the total variance was explained in the first PC. Therefore, we tried to reduce the offset. In the first study (Chapter 2), we subtracted the mean angle of the static reference trial. However, this was not an option for the follow-up measurements. With the supramalleolar osteotomies, the alignment of the hindfoot was changed. Hence, referencing to the static position could mask some effects of the surgery. Therefore, we decided to reduce the offset by subtracting the mean angle over the gait cycle for each trial (Chapter 5). This was successful in reducing the offset and the PCA captured more features within the first few PCs that explained 70% of the total variance. However, with this method the values of the joint angles did not represent anatomical values anymore.

1.1.2 Data analysis

PCA has been used in this thesis to analyze the waveform data of the different joint angles, joint moments, ground reaction forces, or EMG envelopes (Chapters 2 and 5). These waveform data that were obtained from the gait analysis provide a lot of information that often needs to be reduced for the data analysis. Here, PCA provides a powerful method that

achieves a data reduction without any a priori selection of the desired parameters such as a minimal or maximal value. The data reduction with the PCA was achieved by only retaining the first few components that explained up to 70% (Chapter 5) or 90% (Chapter 2) of the total variance. This is a simple method and works well since the PCs are ordered according to the percentage of the total variance that they explain [23]. However, there is no rule on the selection of the cut-off percentage. Generally, the cut-off percentage will be lower with more observations, but one might have to adjust the cut-off percentage according to the desired result of the PCA. If the PCs explain only a small amount of the total variance and data reduction is the purpose of the PCA then a lower cut-off percentage might be useful. On the other hand, if there are one or two very dominant PCs that explain very obvious features, a cut-off percentage of more than 90% of the total variance might resolve more features [23].

The support vector machine (SVM) is increasingly used in the classification of gait patterns of different subject groups, including healthy subjects and patients [24-26]. Both linear and non-linear classifiers are frequently used. In this thesis we used a linear SVM classifier (Chapter 2), because it allows the calculation and thus also the interpretation of the separating hyperplane. However, it is possible, that non-linear kernels (polynomial or radial basis function) lead to better results especially if the data are not linearly separable [27,28]. In this thesis (Chapter 2), the classification was performed with only a few subjects. Although that it was successful, the inclusion of more subjects could lead to a more robust classifier that has a better generalization performance and could be used as a diagnostic tool.

Wavelet analysis and thus wavelet patterns of surface EMG signals have frequently been used to study muscle activation as they allow studying both time and frequency contents of the EMG signal together [29,30]. To visualize differences in the wavelet patterns of various subject groups, difference wavelet patterns were calculated [31,32]. However, the differences were with this method still only visually depicted and not quantified. In this thesis, the entropy was introduced as measure of the homogeneity of the wavelet patterns during one gait cycle (Chapter 3). The entropy is maximal when the intensities at each time point and frequency are equally probable, thus have the same value, while it is zero when all intensities except one have a probability of zero [33]. It indicates how peaky a wavelet pattern is. However, as the entropy is calculated by summing over both the time and frequency axis of the pattern, it doesn't yield information on the location of these peaks. The interpretation of the results on the entropy as broader activation regions in Chapter 3 was thus only possible together with the visual inspection of the patterns and a temporal analysis of the peak muscle activation.

2. Outlook

While this thesis identified gait patterns of patients with asymmetric ankle OA, open questions remain, especially regarding the influence of the muscle activation. Chapter 3 indicated differences in the intermuscular coordination between the measured lower leg muscles. Patients with a valgus alignment of the hindfoot had an altered timing between *Mm. gastrocnemius medialis*, *lateralis* and *soleus* peak activity and different wavelet patterns of *M.*

peroneus longus (Chapter 3). However, the number of studied patients with a valgus alignment was too small to draw a conclusion on whether that was particular for these patients or whether this was general adaptation in patients with ankle OA and a valgus hindfoot alignment. For future studies it would be interesting to see whether this different coordination of the gastrocnemii is generally seen in patients with valgus ankle OA and whether also healthy subjects without OA but with a valgus alignment of the hindfoot show such changes.

The analysis of the gait patterns by PCA and the subsequent classification by SVM (Chapter 2) could be further extended. Instead of only using kinematic and kinetic data, one could also use EMG data such as EMG envelopes or even wavelet patterns. With more prospective data, it would be interesting to see whether the patients are classified as healthy after realignment surgery. Further, a larger data set of gait patterns of ankle OA patients could answer whether the preoperative gait pattern influence the outcome of the surgery. Supramalleolar osteotomies do not exclude further treatment of the ankle OA with either ankle arthrodesis or total ankle replacement in case of persisting pain and progressing OA. Two patients that participated in the prospective part of this thesis were not available for a follow-up gait analysis as their OA progressed and they had either an ankle arthrodesis or a total ankle replacement within one year after realignment surgery. Hence, it would be interesting to see, if there were already indications in the gait pattern of these two patients that could predict a further OA progression.

3. Conclusion

Several pattern recognition methods such as PCA, SVM, and wavelet analysis together with conventional analysis methods such as reporting ranges and peak values were successfully applied in this thesis for the objective quantification of the patients' gait patterns. The biomechanical adaptations in the gait patterns included reduced ranges of motion and reduced peak kinetics (i.e. ankle dorsiflexion moment) and were mainly related to a reduced mobility of the ankle joint. Further, neuromuscular adaptations such as changes in the frequency distribution and the intermuscular coordination were found. Contrary to the biomechanical adaptations, the reasons for these changes and their role in the rehabilitation process are less understood and need further investigations. Finally, the main result of this thesis was that while the gait patterns after joint preserving realignment surgery resemble the preoperative ones, they are comparable to those of patients with an ankle arthrodesis or a total ankle replacement, but the pain level decreases and the general well being and quality of life increases.

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Acknowledgement

At last, my years as a PhD student are coming to an end. It was not always as straightforward as I wanted, but thanks to the encouragement and support from many people I could successfully write my thesis.

First, I would like to thank my supervisor Prof. Victor Valderrabano for the possibility to write my PhD thesis in his research group. You always supported and believed in me. You helped with the interpretation of my data with both your clinical experience and biomechanical knowledge and challenged me with many ideas for new papers. My thanks go also to Dr. Geert Pagenstert. You motivated me and helped with organizing the preoperative gait analysis, preparing presentations, or correcting abstracts and manuscripts.

I'm grateful to my mentor Dr. Vinzenz von Tscharner. Thank you for sharing your vast experience in signal processing, especially for EMG signals with me. This helped me with the analysis and interpretation of EMG data. You also encouraged me to not be satisfied with my results and always think of different ways to analyze my data.

I thank the Swiss National Science Foundation for the funding of this PhD project.

I would like to thank Prof. Philippe C. Cattin for the discussions and his input in pattern recognition methods and Beat Göpfert for his help with data collection and solving technical problems. Further, my thanks go to the team of the gait laboratory of the University Children's Hospital Basel for their support, to all the doctors from the University Hospital Basel that helped me recruiting patients for my studies, and to Dr. Alexej Barg for his help with writing the clinical paper. I am also thankful to all patients and other participants that took their time to participate in this study.

But life is not only about work. So, I would like to thank my friends and fellow PhD students Cora, Sarah, and Katrin for sharing the ups and downs of the life as a PhD student. It was a pleasure spending time together during lunch, at conferences, and going hiking or swimming in the Rhine. Thank you also to all the other people that shared the office with me and that had time for talks and coffee breaks.

A big thank you goes also to my teammates from the floorball teams I played in during my PhD years. I love the sport and being part of a team. The practices and games were the best way to get my mind off the thesis and focus on something completely different. I am also very grateful to my friends. Thank you for going hiking or skiing, for chatting, or for going out for a drink.

Last, but not least, I want to thank my family, especially my mother and my brother. You were always there for me and supported me throughout my life.

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