© Med Sci Monit, 2008; 14(3): CR159-169
PMID: 18301360

CR

Received: 2008.02.06
Accepted: 2007.02.07
Published: 2008.03.01

Changes in gait parameters in total hip arthroplasty patients before and after surgery

Benedykt Cichy[1ABCDEFG], **Magdalena Wilk**[2,3ADEF], **Zbigniew Śliwiński**[3DE]

[1] Traumatology, Orthopedics and Rehabilitation Clinic, College of Medicine, Jagiellonian University, Cracow, Poland
[2] Cracow Rehabilitation Center, Cracow, Poland
[3] Swietokrzyska Academy, Kielce, Poland

Authors' Contribution:
A Study Design
B Data Collection
C Statistical Analysis
D Data Interpretation
E Manuscript Preparation
F Literature Search
G Funds Collection

Source of support: This study was financed in part by the Polish Scientific Research Committee (KBN), grant no. 3 P05 D 070 23

Summary

Background: The ideal outcome in total hip arthroplasty (THA) with endoprosthesis is the elimination of pain and the recovery of a normal range of movement in the affected hip joint, which is essential in order to improve the gait and restore the quality of life. In rehabilitation programs aimed at teaching patients to walk after THA, it is of particular importance to restore proper gait rhythm, speed, and fluidity of motion.

Material/Methods: We examined 30 patients with degenerative changes of the hip joint (11 men, 19 women), who had been referred for THA in the period 2002–2004 due to unilateral degeneration of the hip joint. Pedobarography was used to record the distribution of force on the foot in each patient just before and again one month after surgery, along with clinical tests to measure the range of motion (ROM) for both lower extremities. The body mass index (BMI) was also measured.

Results: Static measurements showed that before surgery there was no statistically significant asymmetry between the affected and healthy lower limbs in respect to maximum foot-ground pressure. One month after surgery, however, we found some asymmetry, caused by reduced load on the operated limb. After THA there was a slight increase in step length in both limbs, but asymmetry in step length persisted.

Conclusions: One month after THA with endoprosthesis we observed slight improvement in step length and increased asymmetry in load-bearing in the affected limbs.

key words: **pedobarography • endoprosthesis • osteoarthritis • asymmetry of gait**

Full-text PDF: http://www.medscimonit.com/abstract/index/idArt/836592

Word count: 6670
Tables: 4
Figures: 1
References: 45

Author's address: Magdalena Wilk, Cracow Rehabilitation Center, Modrzewiowa 22, 30-224 Kraków, Poland

Background

Walking is the most essential modality of human locomotion, and thus disturbances of gait have a significant impact on the quality of life. For the present purposes, gait can be defined as the rhythmical, alternating movement of the lower limbs, combined with shifting the body's center of gravity in the desired direction. During walking one of the feet is always in contact with the ground; this is what distinguishes walking from running, which includes a leap phase. In order to walk it is necessary to use all the joints of the lower limb in an extraordinarily complex movement chain [1]. Moreover, an essential role is also played by spinal mobility (including the cervical spine) and alternating movements of the upper limbs [2,3].

The participation of virtually every part of the body in the mechanism of gait requires efficient control by the nervous system, which explains why walking is not possible immediately after birth in humans (as opposed to many animals), due to the immaturity of the infant's nervous system. Physiological gait is an extremely energy efficient form of locomotion, which means that any disturbance of its normal mechanisms is accompanied by increased energy costs and decreased muscle efficiency. Thus when patients in rehabilitation must learn to walk again, it is essential to conduct a thorough evaluation of the existing movement abnormalities in the joints and the functional parameters of the various muscles and muscle groups.

The kinematic pattern of the normal movements that take place in all the joints involved in walking is similar in the sagittal plane in nearly all individuals [2,4]. Inter-subject differences can be seen in kinematic analysis primarily in the frontal projection (e.g. some people sway more from side to side while walking normally than others) [1,2]. The forces that operate while walking are generated by muscle actions that accelerate or retard the movement of various body segments, gravity, and momentum.

Gait is often described as a series of controlled falls that are not completed, due to the reflex movements that restore equilibrium. This definition, though it seems oversimplified, is nevertheless very close to the truth of the matter [5]. Taking the first step involves throwing the foot forward and shifting the point of gravity in relation to the foot, which entails a loss of balance in the forward direction. The force of gravity and the momentum generated by this motion continue the forward movement of the entire body. Falling is avoided by the reflex reaction of planting the foot, which cause the center of gravity to be shifted back to the inside of the quadrilateral of ground support, i.e. between the feet. If the walk is to be continued, the center of gravity must once again be shifted forward. By the second step propulsion is now possible with much less contraction of the flexor muscles in the calf of the leg, thanks to the effect of the momentum acquired during the first step; the body is shifted forward and the next steadying reflex is evoked. This cyclical mechanism is continued for as long as necessary, while the acquired momentum enables energy savings as soon as an even cadence of steps is achieved. There are only two moments when walking on a level surfaces requires a greater energy expenditure: when starting to walk (since it is necessary first to overcome the force of inertia in order to move the body forward) and again when stopping (since the movements of the limbs and trunk propelled by momentum must be resisted) [5,6].

Any disturbance of this extraordinarily efficient mechanism for exploiting the force of momentum will cause a loss of energy efficiency in the gait [2,4]. Therefore, when we are teaching patients to walk again, it is essential to restore the normal rhythm of gait, including the fluid movement of both lower and upper limbs [7,8].

One of the most common dysfunctions that can lead to gait disturbances is osteoarthritis, especially in the hip, leading to pathological changes that considerably impede gait efficiency and slow it down. According to the definition proposed in 1995 by Keuttner and Goldberg, osteoarthritis is the result of both biological and mechanical destabilizing processes that involve the degradation and abnormal synthesis of articular cartilage at the level of the chondrocytes and the extracellular matrix, as well as the subchondral layer of the bone [9]. This pathological process can be initiated by many different factors. As it progresses, the degenerative process involves all the tissues in the joint, and manifests itself as morphological, biochemical, molecular, and biomechanical changes in the matrix cells, which lead to softening, fibrosis, ulceration, and wasting in articular cartilage, along with hardening and thickening of the subchondral tissue of the bone, and the formation of osteophytes and subchondral cysts. The hip joint is a frequent target of osteoarthritis, which affects ca. 10–20% of the white population, and 85% of persons over age 75 [10].

The treatment of choice for osteoarthritis of the hip joint is presently total hip arthroplasty (THA). This operation, when properly conducted and not attended by complications, is one of the most spectacular achievements of modern orthopedic surgery. Successful treatment outcome consists in completely eliminating or significantly reducing pain while restoring or increasing the range of motion (ROM) in the hip joint, which in turn enables restoration of normal gait and increased quality of life [3,11,12]. At present total endoprostheses are in use either without cement (when the endoprosthesis is implanted in the bone using bone cement) or with cement (implanted on a bed of bone cement). Until fairly recently it was generally assumed that cementless endoprostheses were indicated in patients who were not above a certain age limit, generally 60–65. Presently, however, most authorities believe that the use of one or the other type of endoprostheis should depend primarily on the state of the bone tissue in the given patient; the clinical outcomes and useful life of the implants do not vary significantly between cementless and cemented endoprostheses [13]. The use of cementless endoprostheses is now considered by most authorities in the field to have more advantages, since hip revision surgery, if required, is technically simpler without the cement.

Terminology and research methodology in studies of gait

The mechanical complexity of the speciously simple process of walking requires that we break the process down into particular segments in order to analyze gait. Some of the terms used refer to the duration of certain segments, while others involve spatial relations, physical forces, and the intervals through which anatomical structures must move. Gait disturbances are manifested both temporally and spatially, which means that both aspects must be taken into account [14].

One of the parameters commonly used is step length, understood as the distance between the right and left footprints while walking (ca. ¾ of a meter in most healthy adults), as opposed to the length of the walk cycle, which is the distance covered by both lower limbs in two successive steps (ca. 1½ meters in most healthy adults).

The next parameter is the duration of the walk cycle, i.e. the time it takes to complete one full sequence of movements. This can be measured by noting the time that elapses from the first contact of the right foot with the ground to the next contact of the right foot with the ground.

The walk cycle itself is divided into two phases:
- the stance phase, when the foot is in contact with the ground;
- the swing phase, when the foot is being moved forward preparatory to making the next step.

During walking there is also a certain period of time when both feet are in contact with the ground at the same time: i.e. the rearward limb is still in contact with the ground with the toes and forefoot, while the heel of the forward limb has already made contact with the ground. This overlap period is called the "double stance phase." While walking at medium speed, the stance phase occupies about 60% of the walk cycle, and the swing phase about 40% [4]. During a slow walk, the stance phase is relatively prolonged, and can amount to as much as 70% of the entire gait cycle. As gait speed increases, the duration of the stance phase is slightly decreased, and during very rapid walking can fall to less than 57% of the walk cycle [5]. The duration of the double stance phase is also shortened as the speed of walking increases. In a very slow walk, the double stance phase may last up to 46% of the entire walk cycle, while at a very quick walk it is only 14%.

Speed and cadence are also important gait parameters. Speed, i.e. the distance covered in a specified unit of time in physiological conditions, is usually ca. 1.5 m/sec. (5 kmh) in a healthy adult. Cadence is the number of steps made by the right and left lower limb in one minute (ca. 110 steps per minute in most adults). As in the motion of a pendulum, the lower limb moves with a cadence that is inversely proportional to the length of step, which is why the gait cadence of shorter individuals is generally higher than that of taller persons. In persons over age 65 there is a tendency to shorten the walk cycle, but the cadence remains basically unchanged. If a stop watch is used to measure speed while simultaneously counting the number of steps needed to cover a given distance, the following equations can be used to calculate step length in clinical conditions:

$$\text{gait speed} = (\text{length of step} \times \text{cadence})/120,$$

thus

$$\text{length of step} = (120 \times \text{speed})/\text{cadence}.$$

Gait speed is very sensitive to pathology, thus many authors emphasize the importance of measuring this parameter during rehabilitation in order to assess progress [15–17].

Gait can also be evaluated by using several physical determinants, i.e. mechanisms that are used by the body to minimize the shifting of its center of gravity. The kinematic determinants include the following:
- the turn of the pelvis in the frontal plane;
- the slant of the pelvis in the frontal plane;
- knee flexion during the stance phase;
- dorsal flexion and extension in the upper ankle joint during the stance phase;
- knee flexion at the moment when the foot is being prepared to push off;
- alternating shifts of the pelvis in the horizontal plane.

The temporal gait determinants that can be used to characterize normal gait are as follows:
- isometry, when the steps made with both lower limbs have the same length;
- isotony, when the movements of the upper and lower limbs while walking are properly coordinated;
- isochronicity, when the duration of weight bearing on both lower limbs is equivalent.

The disturbances caused by pathological processes in even one of these determinants can increase the shifting of the point of gravity while walking, at the cost of the energy efficiency characteristic of normal gait [2].

The movement pattern in the hip while walking is considerably less complex than in the knee or ankle joints. In the entire walk cycle, the hip joint has one extension phase and one flexion phase, whereas the knee and the ankle each make two phases of each kind of movement during one cycle. In the knee and upper ankle joints, however, the range of motion is mostly limited to the sagittal plane (flexion and extension), whereas a free range of motion in all three planes is essential for the proper functioning of the hip joint in normal gait. When the heel strikes the ground, the hip joint is in slight flexion, while the gluteus maximus and the posterior thigh muscle group immediately contract, in order to initiate the extension of the hip joint. The knee is fully extended or flexed ca. 5°, while the posterior thigh muscle group controls the flexion of the knee that occurs after the heel touches the ground. The upper ankle is in full dorsal flexion. The body mass shifts onto one lower limb, and the thigh adductor muscles are engaged, followed almost immediately by the thigh abductors, stabilizing the pelvis in relation to the thigh. At the same time the gluteus maximus tightens, straightening the hip and inhibiting the inward rotation of the thigh. During the stance phase there begins an extension in the hip joint due to the action of the extensors, and the knee increases its flexion in order to minimize the effect of the heel striking the ground and the horizontal shift of the center of gravity, which occurs when the body weight is shifted forward over the limb that has been stabilized on the ground. The flexion of the knee during the stance phase is one of the classic determinants of gait, and can reach 30°. In the ankle, there is controlled plantar flexion in order to allow the foot to land safely on the ground.

In dynamic testing, the period of time when the foot bears weight during the stance phase is known as the propulsion phase, sometimes called the "switching phase" or the "rocking phase." Tension in quadriceps softens the impact of the

CR

heel on the ground and controls the momentum of the body mass, now pushing the knee forward. In the stance phase there is virtually no muscle activity outside of the calf muscles, which begin to act during this phase. They achieve their highest level of activity just before push-off, when the heel is lifted from the ground due to the shift of the center of gravity to the forefoot after full dorsal flexion of the foot has been achieved. In the final period of this phase the contraction of the foot flexors adds the motion component necessary for push-off. Then once again the thigh adductors are engaged and the cycle begins all over again [2,18].

A full gait analysis includes testing the strength of foot pressure on the ground, three-dimensional video recording of the motion of the patient's anthropometric points and electromyographic (EMG) tests of the activity of the muscles that are involved in walking. All three of these methods should be used to obtain the fullest possible picture of gait disturbances, but in clinical practice this is seldom done. The reason for this is the high cost and inaccessibility of the research apparatus involved, as well as the discomfort the patient must endure to go through all these tests. The subject must be nearly completely undressed for a long period of time: the preparations for the procedure involve a long and tiresome process of fastening markers to the patient's body at the selected anthropometric points, and the even more laborious procedure of mounting electrodes for the EMG, so that the whole testing procedure can take several hours [14,15,18]. This methodology, though it can certainly provide much valuable data, seems highly impracticable in the case of patients with osteoarthritis of the hip joint, since these are often older persons with pronounced limitations of gait, as well as pain and discomfort when moving.

The technique of measuring the force exerted by the body on the ground uses Newton's third law of dynamics: the action of the forces exerted by a person walking evokes an equal but opposite reaction. It is precisely the ground reaction force that exerts an effect diminishing the shift in the center of gravity during the walk cycle [2]. Several different methods have been described to measure the force exerted on the ground by the foot. The classic methods, such as plantoconturography or podoscopy can measure only the static values of the pressure (i.e. while standing), which means that they do not enable us to measure changes in force during the various phases of the walk cycle [19]. In more advanced systems, the objective measurement of ground pressure is made using a matrix of piezoelectric or capacitive transducers, connected to a computer that records changes in the forces exerted on particular sensors over time. The most commonly used systems are based on shoe inserts (such as Pedar®, Novel GmbH), useful primarily in the analysis of functional changes occurring in the foot itself [20–23] and in the dynamic analysis of motion patterns in athletes. For the examination of gait pathologies, however, their use entails a significant margin of error resulting from the interaction of the inserts themselves with the foot and the shoe [21,22]. A more precise, though more complicated method involves the use of a platform that records the force exerted by the foot during the entire walk cycle. Such a device consists of a plate built into the floor that contains a matrix of transducers (e.g. Emed-SF®, Novel GmbH). The density of sensors on these platforms is on the order of $4/cm^2$, which makes it possible to measure force on a scale of hundreds of Newtons on each sensor with an assigned frequency, i.e. sampling (successive measurements) made several dozen to several hundred times per second [24–26].

The patterns of changes in foot-ground pressure are extraordinarily sensitive to any pathological changes in gait, such as those characteristic of pain ("antalgic" or "painful" gait), loss of balance, disturbances in neuromuscular control [27,28]. Just as in the case of the kinematic patterns of gait analysis (which involves changes in the spatial positioning of the orientation points in the body and the activity of the various muscle groups), it is not easy in pedobarographic examinatinons to obtain a typical pattern of normal gait that is similar in all subjects. Due to the randomness of some movements, there is often a large dispersion that complicates the comparison of data obtained from different patients, given the anatomical differences in the structure of the foot, the differences in force caused by differences in body weight, and individually variable differences in the way one moves [29]. Thus it is advisable to perform a comparative analysis for several (preferably 3–5) examinations and calculate the arithmetic mean of the results [20,30,31]. In analyzing the data one should also take into account the distribution of forces in the anatomical regions of the foot, and draw up graphs of the changes taking place during the walk cycle. When the distribution of force is known, as well the changes that take place over time, it is possible to make the necessary calculations for the standardized anatomical areas, known in the literature (and hereinafter) as "masks" [26,32,33].

Material and Methods

Our research was performed in the Traumatology, Orthopedics, and Rehabilitation Clinic of the College of Medicine at the Jagiellonian University in Cracow, Poland, and at the Central Footwear Industrial Laboratory, also in Cracow, in the period 2002–2004. We examined a group of 30 patients, 11 men and 19 women, ranging in age from 50 to 80 years (mean 63.6±8.9 years). The average age of the men was 62.5±8.9 years, which was not significantly different from that of the women (64.3±9.2 years). The inclusion criteria were as follows:

- unilateral, primary osteoarthritis of the hip joint;
- BMI below 30;
- preserved capacity for unassisted locomotion.

The exclusion criteria included:

- a history of disturbances of arterial or venous circulation in the lower limbs;
- a diabetes;
- clinically manifest degenerative changes, limited range of motion, or traumatic changes in the other lower limb or spine;
- lower limb amputation;
- one lower limb more than 2.5 cm shorter than the other;
- deformation or disease in one or both feet.

All these patients underwent THA with endoprosthesis in our clinic. In 16 patients a cemented endoprosthesis was used (KERAMED®, Mathys, Switzerland), while in the remaining 14 cases a SAMO® cementless endoprosthesis was implanted (Samo SpA, Italy). All patients were operated using an anterior-lateral approach in epidural anesthesia.

The same rehabilitation regime was applied in all cases. The program, which has been used in our clinic since 1995, calls for early load-bearing on the operated limb and early implementation of unassisted walking. Rehabilitation, which begins on the second day after surgery, includes active exercises for the healthy lower limb in the foot, knee and hip, general fitness and isometric exercises for the upper limbs, respiratory exercises, and passive exercise of the operated limb, using an ARTROMOT K-2 device (Ortmed GmbH, Germany) to provide continuous passive motion (CPM). The patient is first brought to vertical position while sitting on the bed with the lower limbs hanging over the edge. Full standing and learning to walk with partial loading of the operated limb (using a walker) is commenced on the third day. In the period from the fourth to the seventh day, supported or active exercises for the operated limb are introduced in all joints (with the exception of strong adduction and internal or internal rotation in the operated hip); the patient learns to walk with two elbow crutches. On the 7th day after surgery the patients continue the previous exercises, and then gradually begin to learn to walk with increasing load on the operated limb, to walk with one elbow crutch, to walk without assistance, and to go up and down stairs. This program is continued until the patient is discharged from the hospital. Along with their discharge documentation, the patient receive a set of home exercises to increase muscle strength and range of motion in the operated hip joint. Particular attention is paid in our program to restoring patient independence as soon as possible, and to attaining normal gait. The patient is encouraged to put full weight on the operated limb as soon as that can be done safely, and to give up the security of crutches and canes as far and as soon as possible. The good effects achieved thanks to this program have been confirmed in several publications from our clinic [34,35]. This is consistent with the current literature, which stresses early rehabilitation and the effort to restore natural movement [7,8].

Each of the patients was examined twice. The first examination was performed before the surgery, typically 1–7 days before the planned date of admission to the orthopedic surgery ward. The second examination was performed one month after surgery.

For gait evaluation we applied static and dynamic testing of the distribution of force recorded on the sole of the foot using an EMED SF-4 device (Novel, Munich, Germany, cf. Figure 1). The work station consists of a fixed runway 5 meters long and 1 meter wide, with a 360×190 mm platform built in at ca. 2/3 of the length, used to measure dynamically forces from 0 to 127 N [26].

The protocol for each examination was identical. The tests on the pedobarographic platform were performed three times, using the method described in the literature as midgait measurement [24,30,31,36]. In this method the measurement is taken for the second step after the stance phase, which considerably reduces error resulting from the momentary loss of balance observed when a person begins to walk. The "first step" method occasionally encountered in the literature is applied primarily in tests of persons with a major disability or with significant disturbances of equilibrium; the advantage of this method is that it is much easier to administer in such persons, but it is subject to error,

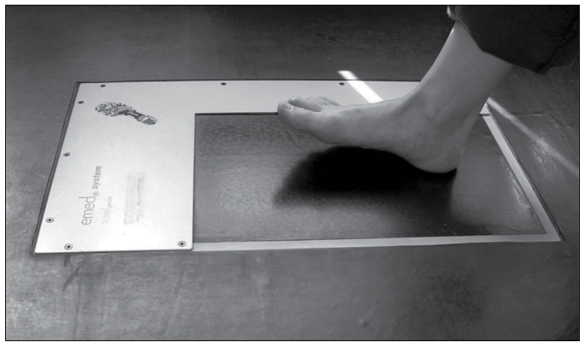

Figure 1. EMED SF-4 pedobarographic platform for the dynamic measurement of force on the sole of the foot.

as previously mentioned, and is thus generally regarded as much less exact [24,25,30,31,37].

The data acquired during static and dynamic testing were then analyzed using a dedicated computer program. The image of the foot was overlaid with seven masks, dividing the foot into specific anatomical areas:

- Mask no. 1: the big toe;
- Mask no. 2: the remaining toes;
- Mask no. 3: the head of the first metatarsal;
- Mask no. 4: the heads of the second and third metatarsals;
- Mask no. 5: the heads of the fourth and fifth metatarsals;
- Mask no. 6: the midfoot;
- Mask no. 7: the heel [23].

This system of masks enables us to obtain standardized force values for the various areas of the foot [16,24,32]. The masks are drawn in a special computer program and recorded as a separate file for each patient. Once saved to disk, the masks can then be reused for the same subject in later examinations, since it is rarely necessary to make more than minor adjustments to impose the existing masks on the new results [33].

Cumulative data were obtained in three sets for each foot in each of the two examinations (before and after). The parameters for each foot in one series were averaged (arithmetical mean), so that for each examination only one result is given. This procedure is generally accepted in the literature [38].

Clinical tests were also performed on al patients, measuring the range of motion in the hip joint and the length of the lower limbs; the length of step was measured three times in succession for each lower limb, to obtain the arithmetical mean from the three values obtained. In addition, during the first examination we measured the patient's height and weight to calculate the body mass index (BMI).

We were looking for changes taking place in the gait mechanics, primarily in such parameters as:

- step length;
- foot-ground contact time;
- total area under the time-force curve;
- the moment of occurrence of maximum force as a percentage of the total duration of the propulsion phase.

The purpose of adopting this method was to evaluate the symmetry or asymmetry of gait. All the parameters were compared for the affected and healthy legs, separately for each examination (before and after surgery). The criterion for improvement in gait was assumed to be the restoration of symmetry in load bearing on the lower limbs (though the results indicated asymmetry) [17,38,39].

All the data obtained from both examinations were subjected to statistical analysis, calculating the arithmetical mean (x), standard deviation (SD) and range (minimum-maximum) for each parameter. Since in many cases the distributions were non-normal, non-parametric methods were used. To compare the mean values for the affected lower limb (ALL in the figures and tables) to the means for the healthy lower limb (HLL) in the same patients we used the non-parametric Wilcoxon test for matched pairs [40]. In examining endoprosthesis type (cemented or non-cemented), since we were comparing results in two different groups of patients, the non-parametric U Mann-Whitney test was used [40]. The statistical hypotheses were verified assuming $\alpha=0.05$ as the level of statistical significance. The Statistica PL 6.0 statistical software package (StatSoft, Tulsa, Oklahoma, USA) was used, applying the modules for "Basic statistics and tables" and "Non-parametric statistics."

RESULTS

Static testing

We first analyzed the maximum force as recorded for both the affected and healthy lower limbs in standing position. In the first examination, we found a mean maximum force of 17.2 N/cm^2 in the healthy limb, as compared to 14.4 N/cm^2 in the affected limb. In the second examination, one month after surgery, the results were 15.8 N/cm^2 and 13.8 N/cm^2, respectively. No significant effect of gender or type of endoprosthesis was found.

Dynamic testing

Table 1 identifies the parameters where we found significant differences between the affected and healthy limbs; Table 2 breaks this information down according to the particular masks (anatomical regions of the foot).

In the first examination (prior to surgery) we found asymmetry of gait parameters in three masks: no. 1 (big toe), no. 2 (remaining toes), and no. 7 (heel). These differences proved to be highly significant using the Wilcoxon test ($p<0.001$). Table 3 gives the results for these parameters.

In mask no. 1, maximum force, contact time, and the area under the force/time curve were less in the affected limb than in the healthy limb. In mask no. 2, lower values were obtained in the affected limb for maximum ground pressure, area under the pressure/time curve, and area under the force/time curve. In mask no. 7, higher values were observed in the affected limb for the moment of maximum force as a percentage of the propulsion phase.

In the second examination we found considerable more indication of gait asymmetry in all seven masks (see Table 4). The values presented in Table 4 differ significantly ($p<0.001$, Wilcoxon test) between the two lower limbs.

Table 1. Summary of gait parameters in which there occurred statistically significant differences ($p<0.001$, non-parametric Wilcoxon test) between the affected and healthy limbs in either the first or second examination.

Parameter	Exam I	Exam II
Total area of surface contact	↓	↓
Maximum force		
Maximum pressure	↓	↓
Contact time for the entire foot		↓
Area under the pressure/time curve	↓	↓
Area under the force/time curve		↓
Moment of maximum pressure		↓
Moment of maximum pressure as a percentage of propulsion		
Moment of maximum force		
Moment of maximum force as a percentage of propulsion		

↓ parameter is lower in the affected lower limb than in the healthy lower limb;
↑ parameter is higher in the affected lower limb than in the healthy lower limb.

In mask no. 1 the values for the affected limbs were lower than those for the healthy limb in surface area of contact, contact time, area under the force/time curve, moment of maximum ground pressure, and moment of maximum force. In mask no. 2, the same was true for area under the pressure/time curve, area under the force/time curve, moment of maximum ground pressure, moment of maximum force, and moment of maximum force expressed as a percentage of the propulsion phase. In mask no. 3, there were two parameters with lower results for the affected limb (contact time and beginning of contact as a percentage of the propulsion phase), while contact time as a percentage of the propulsion phase gave a higher value for the affected limb than the healthy limb. In mask no. 4 we found lower values in the affected limb for contact time, beginning of contact as a percentage of the propulsion phase, and moment of maximum ground pressure, whereas contact time as a percentage of the propulsion phase was higher in the affected limb. In mask no. 5 the results were lower in the affected limb for contact time and moment of maximum ground pressure. In mask no. 6, only contact time was lower in the affected limb. In mask no. 7, three parameters (maximum ground pressure, contact time, and beginning of contact as a percentage of the propulsion phase) were lower in the affected limb, while moment of maximum ground pressure as a percentage of the propulsion phase was lower in the healthy limb.

Length of step

The average length of step in healthy adults at a freely chosen walking speed is ca. 75 cm, and is equal for both lower limbs. In our patient group, step length measured before surgery was much less than the norm: 42.6±13.1 cm (range

Table 2. Summary of gait parameters in which there occurred statistically significant differences (p<0.001, non-parametric Wilcoxon test) between the affected and healthy limbs in either the first or second examination, broken down by mask (anatomical region of the foot).

Parameter	Mask 1		Mask 2		Mask 3		Mask 4		Mask 5		Mask 6		Mask 7	
	I	II	I	II	I	II	I	II	I	II	I	II	I	II
Total area of surface contact		↓												
Maximum force	↓													
Maximum pressure			↓											↓
Contact time for the entire foot	↓	↓				↓		↓		↓		↓		↓
Contact time as a percentage of propulsion						↑								
Beginning of contact as a percentage of propulsion						↓								↓
End of contact as a percentage of propulsion														
Area under the pressure/time curve			↓	↓										
Area under the force/time curve	↓	↓	↓	↓										
Moment of maximum pressure		↓		↓		↓				↓				
Moment of maximum pressure as a percentage of propulsion														
Moment of maximum force		↓		↓										
Moment of maximum force as a percentage of propulsion				↓									↑	↑

↓ parameter is lower in the affected lower limb than in the healthy lower limb;
↑ parameter is higher in the affected lower limb than in the healthy lower limb.

Table 3. Results of tests in which significant differences occurred in gait parameters between the affected and healthy lower limbs in the first examination.

Parameters	ALL			HLL		
	x ±SD	Min	Max	x ±SD	Min	Max
			Mask no. 1			
Maximum force (N)	91.4±57.9	8.6	209.3	118.6±76.6	8.6	295.4
Contact time (ms)	854.3±497.5	180.0	2080.0	1007.3±498.8	20.0	2020.0
Area under the force/time curve (N×s)	37.8±34.3	1.7	156.1	69.8±86.1	0.2	356.3
			Mask no. 2			
Maximum pressure (N/cm^2)	18.5±14.8	3.5	80.0	23.3±21.4	4.5	123.5
Area under the pressure/time curve (N/cm^2×s)	8.3±6.6	0.6	22.4	11.0±10.3	0.1	52.7
Area under the force/time curve (N×s)	19.9±16.4	0.6	73.4	24.7±20.1	0.1	67.6
			Mask no. 7			
Moment of maximum force expressed as% of propulsion	33.5±11.6	15.5	56.0	28.1±13.7	5.6	54.7

21–70 cm) for the affected limb, as compared to 44.1±13.7 cm (range 21–74 cm) for the healthy limb. After THA, the mean length of step was slightly increased in both limbs: 45.7±12.7 cm in the affected limb, 47.2±13.3 cm in the healthy limb. At the same time, we observed slight asymmetry of step length between the two lower limbs: a minimum of 26 cm for the

Table 4. Results of tests in which significant differences occurred in gait parameters between the affected and healthy lower limbs in the second examination.

Parameters	ALL			HLL		
	x ±SD	Min	Max	x ±SD	Min	Max
Mask no. 1						
Contact time (ms)	792.0±328.5	20.0	1580.0	928.7±260.6	440.0	1420.0
Area under the force/time curve (N×s)	47.1±33.8	0.4	121.3	62.2±47.5	9.5	169.3
Moment of maximum ground pressure (ms)	791.0±280.5	10.0	1550.0	934.3±277.5	330.0	1630.0
Moment of maximum force (ms)	782.0±294.2	10.0	1530.0	933.3±250.1	610.0	1610.0
Mask no. 2						
Area under the pressure/time curve (N×s)	8.9±7.3	0.1	30.0	10.7±7.7	1.1	28.9
Area under the force/time curve (N×s)	20.5±17.5	0.1	77.4	28.2±23.4	1.5	100.0
Moment of maximum ground pressure (ms)	777.7±232.3	10.0	1130	907.6±336.8	16.6	1590.0
Moment of maximum force (ms)	769.0±221.4	10.0	1110.0	951.3±256.8	640.0	1550.0
Moment of maximum force as % of propulsion	77.0±16.1	36.8	95.6	82.5±11.6	44.5	94.3
Mask no. 3						
Contact time (ms)	858.0±279.0	20.0	1500.0	948.3±227.2	650.0	1380.0
Beginning of contact as % of propulsion	8.5±8.1	0.0	44.1	10.4±6.4	0.0	33.9
Contact time as % of propulsion	84.7±11.2	36.8	100.0	82.2±7.5	60.7	92.7
Mask no. 4						
Contact time (ms)	889.3±273.8	20.0	1540.0	1002.3±261.2	670.0	1580.0
Beginning of contact as % of propulsion	5.9±4.4	0.0	14.6	6.2±4.2	0.0	16.0
Moment of maximum ground pressure (ms)	766.9±258.4	10.0	1350.0	830.2±281.0	71.0	1350.0
Contact time as % of propulsion	87.3±6.8	69.1	100.0	86.7±6.0	71.4	96.2
Mask no. 5						
Contact time (ms)	854.0±269.1	20.0	1500.0	966.7±251.3	620.0	1480.0
Moment of maximum ground pressure (ms)	715.7±204.1	10.0	1090.0	806.7±194.3	330.0	1170.0
Mask no. 6						
Contact time (ms)	754.7±73.8	20.0	1400.0	871.0±231.3	600.0	1480.0
Mask no. 7						
Maximum ground pressure (N/cm^2)	21.6±6.8	12.5	41.5	24.7±6.9	14.5	41.0
Contact time (ms)	690.0±247.1	20.0	1320.0	769.0±256.7	340.0	1380.0
Beginning of contact as % of propulsion	0.0±0.3	0.0	1.4	0.3±1.5	0.0	8.1
Moment of maximum ground pressure as % of propulsion	23.9±13.3	1.0	50.0	21.1±11.9	2.7	41.1

affected limb vs. 25 cm for the healthy limb, as compared to maximum values of 70 cm and 73 cm respectively.

Discussion

Static testing was used to measure the maximum pressure exerted by the lower limbs on the ground while standing. In the first examination there were no statistically significant differences in this parameter between the affected and healthy lower limbs. In the second examination, however, there was a significant reduction in the operated limb. Our observation, then, is that in patients with unilateral osteoarthritis of the hip joint there is no asymmetry in respect to maximum ground pressure. One month after surgery, however, there was less weight bearing in the operated limb, leading to asymmetry. In the literature [41,42] attention is drawn to asymmetry of gait in patients with osteoarthritis of the hip joint; just as in our study, no asymmetry in load bearing while standing can be detected in these patients before surgery, unless one of the lower limbs is at least 2 cm shorter than the other. A similar asymmetry can be observed in patients with significant flexion spasticity in the hip, which produces a functional shortening of the limb [42,43].

In dynamic testing we found a number of differences in the tested parameters between the healthy and affected lower limbs and between the first and second examinations.

In terms of total surface contact area there was a small but statistically significant difference between the results for both lower limbs between the first and second examination. The surface area of contact between the foot and the ground is typically highly variably among individuals, and depends on a number of factors, such as body weight, foot size, or possible deformities of the foot (e.g. flat foot). What seems to have research value is not the absolute value of the parameter compared to norms, but rather the comparison of both feet (assuming there is no structural deformity in one foot or the other). Asymmetry in this respect may also be related to difference in the area under the pressure/time curve, and thus to the pressure exerted on the ground during the entire stance phase [21,33,37].

In respect to maximum force, measured for the foot as a whole, there were no statistically significant differences between the limbs. In the second examination, the average result went down in the affected limb and up in the healthy limb, but the differences were not significant. There was, to be sure, a significant difference in this parameter between the men and the women, but this would seem to be a factor of differences in average body weight between the two groups, rather than changes in gait resulting from the surgery. The maximum ground pressure, as opposed to the maximum force, did not differ significantly between the men and the women for either the affected or the healthy limb. A statistically significant ($p=0.0006$) increase in the maximum pressure exerted by the operated lower limb was observed in the second examination as compared to the first. No statistically significant changes were detected, however, in the healthy lower limb.

In the case of contact time between the foot and the ground, we found no statistically significant differences in the first examination between the affected and healthy lower limbs. In the second examination, there was a statistically significant difference between the two limbs: the mean ground contact time, i.e. the duration of stance phase, was significantly shorter for the affected lower limb. A shorter stance phase results from increased gait speed as physical fitness is recovered.

The parameter called "area under the pressure/time curve" describes the total foot-ground force per unit of area in the whole propulsion phase; differences between the affected and healthy lower limbs, then, can be considered a good indicator of asymmetry of weight-bearing during walking [31]. For this parameter, we found an increased mean for the operated lower limb in the second examination. For the healthy lower limb, this parameter did not change significantly between the first and second examinations. Moreover, this parameter revealed significant asymmetry in both examinations, in that the means for the operated limb were lower than for the healthy limb.

The area under the force/time curve is a derivative of the maximum foot-ground force; this parameter is known to show high inter-individual variability [44]. In the first examination, we found no statistically significant differences in this parameter, but significant asymmetry was observed in the second examination, since the means for the healthy lower limb were higher than those for the operated lower limb.

The moment of maximum pressure and moment of maximum force are parameters that can be used to measure the time needed to achieve maximum pressure or force, respectively, during propulsion. Due to high inter-individual variability, and even variability between successive tests performed on the same patient, this parameter is not considered to be of much value in functional gait analysis [23]. In our study, though we had expected to find results pointing to asymmetry in patients with unilateral hip osteoarthritis before surgery, no statistically significant differences between limbs were found. In the second examination, however, there were slight differences between the operated and healthy lower limbs, in that the operated limbs showed lower values for the moment of maximum pressure than did the healthy lower limb; i.e. the time needed to achieve maximum pressure was shorter. In our opinion, the reduction of mean values of both the moment of maximum pressure and the moment of maximum force between the first and second examinations resulted from reduced ground contact time, which in turn is related to increased gait speed one month after THA [45]. These same parameters, when expressed as percentage values of the propulsion phase, show no statistically significant differences between the lower limbs.

Our methodology for the evaluation of gait parameters was adopted based on a compilation of the methods described in the literature [26,32,33]; we looked for characteristic changes in weight-bearing in the individual areas of the foot, possibly associated with osteoarthritis of the hip or with the THA itself. The results we obtained did not indicate any unequivocal differences in the characteristics of dynamic weight-bearing in the stance phase among the patients we studied.

The differences we found between the affected and healthy lower limbs in respect to the particular anatomical regions of the foot can be summarised as follows:

In the first examination, statistically significant differences were observed between the affected lower limb and the healthy lower limb in respect to weight-bearing parameters in the area of the big toe area (lower mean values of maximum force, contact time, and area under the force/time curve), as well as the remaining toes (lower mean values of maximum pressure, area under the force/time curve and area under the pressure/time curve). This difference suggests asymmetrically lower weight-bearing on all toes in the affected lower limb in patients with osteoarthritis of the hip.

In the heel area, statistically greater values were observed in the first examination for the affected lower limb in respect to the moment of maximum force when expressed as a percentage of ground contact time. This difference, which also occurred in the second examination, suggests that the weight-bearing slope in the heel area generally rises faster during initial contact in the affected lower limb than in the healthy limb.

In the second examination, further differences were found for individual anatomical areas of the foot, likewise pointing to asymmetry of weight-bearing between the operated and healthy lower limb. There were no significant differences between the limbs in respect to pressure and force in individual areas; for the operated lower limb, shorter ground contact time was observed in all masks. In four areas, there were also lower values for the moment of maximum pressure for the operated limb.

In the first examination, statistically significant differences in weight-bearing were observed in three of the seven areas and in six of the thirteen analysed parameters. All these values – apart from one that remained unchanged (the moment of maximum force expressed as a percentage of the propulsion phase in the heel mask) – were lower for the affected limb than for the healthy limb. A total of seven significant differences were observed between the lower extremities. In the second examination, however, many more differences occurred between the two limbs, in all seven masks and in ten of the thirteen parameters we evaluated. As in the case of the first examination, most results (with two exceptions) were lower for the affected limb. All these results point to increased asymmetry and irregularity of weight-bearing at one month post-operatively when compared to the preoperative assessment.

In the evaluation of step length, we observed significant step shortening in both lower limbs in patients with osteoarthritis of the hip when compared to the age-matched norms for healthy adults reported in the literature [39]. The mean values of step length for both the affected and healthy limbs were significantly increased in the second examination, i.e. after surgery: 7.3% for the affected limb and 7.0% for the healthy limb.

Statistically significant asymmetry was also found for step length in the first examination. If the step length of the affected lower limb is taken as a point of reference, the step length of the healthy limb averaged 3.52% longer. There was also a significant difference in the second examination, when the step length in the healthy limb was 3.28% greater.

CONCLUSIONS

In an examination carried out one month after total hip arthroplasty, there was only a small improvement of symmetry in step length and increased asymmetry of weight-bearing between the lower extremities. The methodology used in this study can be useful in clinical practice and for monitoring of physical fitness in patients undergoing hip arthroplasty; it can be also be used to evaluate the outcome of surgery and progress in rehabilitation.

REFERENCES:

1. Persson T: A marker-free method for tracking human lower limb segments based on model matching. Intern J Biomed Comp, 1996; 41(2): 87–97
2. Trew M, Everett T: Human movement: an introductory text. New York: Churchill Livingstone; 1997; 155–65
3. McGibbon CA: Toward a better understanding of gait changes with age and disablement: neuromuscular adaptation. Exer Sport Sci Rev, 2003; 31(2): 102–8
4. Murray MP: Gait as a total pattern of movement. Amer J Phys Med, 1967; 46(1): 290–333
5. Smidt GL, Arora JS, Johnston RC: Accelerographic analysis of several types of walking. Amer J Phys Med, 1971; 50(6): 285–300
6. Mills PM, Barrett RS: Swing phase mechanics of healthy young and elderly men. Human Movet Sci, 2001; 20(4–5): 427–46
7. Hesse S, Sonntag D, Bardeleben A et al: [The gait of patients with full weight-bearing capacity after hip prosthesis implantation on the treadmill with partial body weight support during assisted walking and without crutches] Zeitschrift Orthop Grenz, 1999; 137(3): 265–72
8. Sonntag D, Uhlenbrock D, Bardeleben A et al: Gait with and without forearm crutches in patients with total hip arthroplasty. Intern J Rehab Res, 2000; 23(3): 233–43
9. Keuttner KE, Goldberg VM (eds). Osteoarthritic disorders. Rosemont, California, USA: American Academy of Orthopedic Surgeons, 1995
10. Hoaglund FT, Steinbach LS: Primary osteoarthritis of the hip: etiology and epidemiology. J Amer Acad Orthop Surg, 2001; 9(5): 320–27
11. Hartofilakidis G, Karachalios T: Idiopathic osteoarthritis of the hip: incidence classification and natural history of 272 cases. Orthopedics, 2003; 26(2): 161–66
12. Boardman DL, Dorey F, Thomas BJ, Lieberman JR: The accuracy of assessing total hip arthroplasty outcomes: a prospective correlation study of walking ability and 2 validated measurement devices. J Arthropl, 2000; 15(2): 200–4
13. Zimmerman S, Hawkes WG, Hudson JI et al: Outcomes of surgical management of total hip replacement in patients aged 65 years and older: cemented versus cementless femoral components and lateral or anterolateral versus posterior anatomical approach. J Orthop Res, 2002; 20(2): 182–91
14. Lin JH, Chou YL, Ju MS, Sung YT: [Application and evaluation of the VICON system in gait analysis]. Gaoxiong Yi Xue Ke Xue Za Zhi, 1990; 6(1): 6–11
15. Perron M, Malouin F, Moffet H, McFadyen BJ: Three-dimensional gait analysis in women with a total hip arthroplasty. Clin Biomech (Bristol Avon), 2000; 15(7): 504–15
16. Robinson JL, Smidt GL: Quantitative gait evaluation in the clinic. Phys Ther 1981; 61(3): 351–53
17. Aminian K, Rezakhanlou K, De Andres E et al: Temporal feature estimation during walking using miniature accelerometers: an analysis of gait improvement after hip arthroplasty. Med Biol Eng Comp, 1999; 37(6): 686–91
18. Tsvetkova T, Lebedev V, Makarov Y, Kazimirsky V: Early diagnosis of coxarthrosis (in-shoe plantar pressure measurements). Clin Biomech (Bristol Avon), 1997; 12(3): 18–20
19. Barnett S, Cunningham JL, West S: A comparison of vertical force and temporal parameters produced by an in-shoe pressure measuring system and a force platform. Clin Biomech (Bristol Avon), 2001; 16(4): 353–57
20. Barnett S, Cunningham JL, West S: A comparison of vertical force and temporal parameters produced by an in-shoe pressure measuring system and a force platform. Clin Biomech (Bristol Avon), 2000; 15(10): 781–85
21. Burgess S, Jordan C, Bartlett R: The influence of a small insert in the footbed of a shoe upon plantar pressure distribution. Clin Biomech (Bristol Avon), 1997; 12(3): 5–6
22. Ellegast R, Kupfer J, Reinert D: Load weight determination during dynamic working procedures using the pedar foot pressure distribution measuring system. Clin Biomech (Bristol Avon), 1997; 12(3): 10–11
23. Kernozek TW, LaMott EE, Dancisak MJ: Reliability of an in-shoe pressure measurement system during treadmill walking. Foot Ankle Intern, 1996; 17(4): 204–9
24. Hughes J: Reliability of pressure measurements: the EMED F system. Clin Biomech, 1991; 6: 14–18
25. Quaney B, Meyer K, Cornwall MW, McPoil TG: A comparison of the dynamic pedobarograph and EMED systems for measuring dynamic foot pressures. Foot Ankle Intern, 1995; 16(9): 562–66
26. EMED SF-4 User Manual. Munich: Novel Press, 1994

27. Kemoun G, Defebvre L: [Gait disorders in Parkinson disease: clinical description, analysis of posture, initiation of stabilized gait]. Presse Medicale, 2001; 30(9): 452–59
28. Verghese J, Lipton RB, Hall CB et al: Abnormality of gait as a predictor of non-Alzheimer's dementia. NEJM, 2002; 347(22): 1761–68
29. Vela SA, Lavery LA, Armstrong DG, Anaim AA: The effect of increased weight on peak pressures: implications for obesity and diabetic foot pathology. J Foot Ankle Surg, 1998; 37(5): 416–20; discussion 448–49
30. Peters EJ, Urukalo A, Fleischli JG, Lavery LA: Reproducibility of gait analysis variables: one-step versus three-step method of data acquisition. J Foot Ankle Surg, 2002; 41(4): 206–12
31. McPoil TG, Cornwall MW, Dupuis L, Cornwell M: Variability of plantar pressure data: a comparison of the two-step and midgait methods. J Amer Podiatr Med Assoc, 1999; 89(10): 495–501
32. Cornwall M, McPoil T: The effect of foot orthotics on the initiation of plantar surface loading. Clin Biomech (Bristol Avon), 1997; 12(3): 4–8
33. Orlin M, Stetson K, Skowronski J, Pierrynowski M: Foot pressure distribution: methodology and clinical application for children with ankle rheumatoid arthritis. Clin Biomech (Bristol Avon), 1997; 12(3): 17–19
34. Wilk M, Frańczuk B: [Evaluating changes in the range of movement in the hip joint in patients with degenerative changes, before and after total hip replacement.] Ortop Traumatol Rehab, 2004; 6(3): 342–49
35. Wilk M, Frańczuk B: [Analysis of changes in the strength of muscles acting on the hip joint in patients recovering from hip arthroplasty.] Fizjoter Pol, 2003; 3(4): 309–15
36. Hughes J: The clinical use of pedobarography. Acta Orthop Belg, 1993; 59(1): 10–16
37. Bryant A, Singer K, Tinley P: Comparison of the reliability of plantar pressure measurements using the two-step and midgait methods of data collection. Foot Ankle Intern, 1999; 20(10): 646–50
38. Becker HP, Rosenbaum D, Kriese T et al: Gait asymmetry following successful surgical treatment of ankle fractures in young adults. Clin Orthop Rel Res, 1995; 311: 262–69
39. Shkuratova N, Morris ME, Huxham F: Effects of age on balance control during walking. Arch Phys Med Rehab, 2004; 85(4): 582–88
40. Bland M: An introduction to medical statistics. New York, Oxford University Press, 1995
41. Wall JC, Ashburn A, Klenerman L: Gait analysis in the assessment of functional performance before and after total hip replacement. J Biomed Eng, 1981; 3(2): 121–27
42. Yildiz C, Yanmis I, Tunay S et al: Evaluation of unilateral total hip arthroplasty by a quantitative analysis of gait. Acta Orthop Traumatol Turc, 2002; 36(1): 58–62
43. Lafuente R, Belda JM, Sanchez-Lacuesta J et al: Quantitative assessment of gait deviation: contribution to the objective measurement of disability. Gait Posture, 2000; 11(3): 191–98
44. Willson JD, Kernozek TW: Plantar loading and cadence alterations with fatigue. Med Sci Sport Exer, 1999; 31(12): 1828–33
45. Gotze C, Sippel C, Rosenbaum D et al: Objective measures of gait following revision hip arthroplasty: first medium-term results 26 years after surgery. Zeitschr Orthop Grenz, 2003; 141(2): 201–8