ELSEVIER

Journal of Biomechanics 34 (2001) 883–893

JOURNAL OF BIOMECHANICS

www.elsevier.com/locate/jbiomech
www.JBiomech.com

Musculo-skeletal loading conditions at the hip during walking and stair climbing

M.O. Heller[a,b], G. Bergmann[c], G. Deuretzbacher[d], L. Dürselen[b], M. Pohl[b], L. Claes[b], N.P. Haas[a], G.N. Duda[a,*]

[a] *Trauma and Reconstructive Surgery, Research Laboratory, Humboldt - University of Berlin, Augustenburger Platz 1, Charite, Campus Virchow-Klinikum, 13353 Berlin, Germany*
[b] *Institute for Orthopaedic Research and Biomechanics, University of Ulm, Germany*
[c] *Biomechanics Laboratory, Oskar-Helene-Heim, Free University of Berlin, Germany*
[d] *Clinical Biomechanics, University Hospital Hamburg-Eppendorf, Germany*

Accepted 22 February 2001

Abstract

Musculo-skeletal loading plays an important role in the primary stability of joint replacements and in the biological processes involved in fracture healing. However, current knowledge of musculo-skeletal loading is still limited. In the past, a number of musculo-skeletal models have been developed to estimate loading conditions at the hip. So far, a cycle-to-cycle validation of predicted musculo-skeletal loading by in vivo measurements has not been possible. The aim of this study was to determine the musculo-skeletal loading conditions during walking and climbing stairs for a number of patients and compare these findings to in vivo data.

Following total hip arthroplasty, four patients underwent gait analysis during walking and stair climbing. An instrumented femoral prosthesis enabled simultaneous measurement of in vivo hip contact forces. On the basis of CT and X-ray data, individual musculo-skeletal models of the lower extremity were developed for each patient. Muscle and joint contact forces were calculated using an optimization algorithm. The calculated peak hip contact forces both over- and under-estimated the measured forces. They differed by a mean of 12% during walking and 14% during stair climbing.

For the first time, a cycle-to-cycle validation of predicted musculo-skeletal loading was possible for walking and climbing stairs in several patients. In all cases, the comparison of in vivo measured and calculated hip contact forces showed good agreement.

Thus, the authors consider the presented approach as a useful means to determine valid conditions for the analysis of prosthesis loading, bone modeling or remodeling processes around implants and fracture stability following internal fixation. © 2001 Elsevier Science Ltd. All rights reserved.

Keywords: Hip joint; Loading; Joint contact; Muscle forces; Simulation

1. Introduction

Musculo-skeletal loading plays an important role in the biological processes of fracture healing (Claes et al., 1998), bone modeling and remodeling (Frost, 1999), and in the primary stability of implants (Cheal et al., 1992). Nevertheless, current knowledge of musculo-skeletal loading is still limited. While there is strong evidence that muscles are major contributors to femoral loading (Duda, 1996; Duda et al., 1998; Rohlmann et al., 1983), the actual forces occurring in vivo are hardly accessible.

To date, non-invasive measurement of in vivo muscle forces is still impossible. Ethical considerations discourage the use of invasive methods to determine muscle forces in humans. Therefore, the only opportunity to estimate the complex distribution of muscle forces is offered by computer analysis. In a number of studies, optimization algorithms were employed to solve the distribution problem and simulate loading conditions at the hip (Brand et al., 1986; Collins, 1995; Davy and Audu, 1987; Fuller and Winters, 1993; Glitsch and Baumann, 1997; Herzog, 1987; Pedersen et al., 1987; Pedersen et al., 1997; Röhrle et al., 1984; Seireg and

*Corresponding author. Tel.: +49-30-450-59079; fax: +49-30-450-59969.
E-mail address: georg.duda@charite.de (G.N. Duda).

PII: S0021-9290(01)00039-2

Arvikar, 1973, 1975; Siebertz and Baumann, 1994). A common approach to validating these models was to compare muscle activation patterns obtained from simulation with measured muscle activities as determined by electromyography (EMG). However, this method does not allow quantitative validation of the musculo-skeletal loading conditions. Instrumented implants provide hip contact forces for different activities for individual patients in vivo (Bergmann et al., 1988, 1993, 1995; Brand et al., 1989; Davy et al., 1988; English and Kilvington, 1979; Rydell, 1966). An additional method of validating the predicted musculo-skeletal loading conditions is to compare the calculated hip contact forces with the in vivo measured forces. This comparison will make it possible to determine whether the calculated results are within the range of those found in in vivo studies. The study of Lu et al. (1998) presented a model of the lower limb in the sagittal plane which was validated by a cycle-to-cycle comparison of predicted axial forces in the femur and in vivo forces measured by a massive femoral prosthesis. However, they did not investigate the loading conditions at the hip. To our knowledge, computed hip contact forces and those measured in vivo in the same patient have only been compared in one study (Brand et al., 1994). In this study, hip contact forces were measured 58 days post-operatively, while gait analysis was performed 90 days post-operatively. Therefore, a cycle-to-cycle comparison of measured and calculated hip contact forces was not possible.

The goal of the present study was to quantitatively validate the loading conditions at the proximal femur as predicted by a musculo-skeletal model using a cycle-to-cycle comparison of measured and calculated hip contact forces. In order to minimize the restrictions when dealing with a single activity in an individual patient only, the study was designed to investigate the loading conditions for two typical daily activities in a number of patients.

2. Materials and methods

2.1. Patients

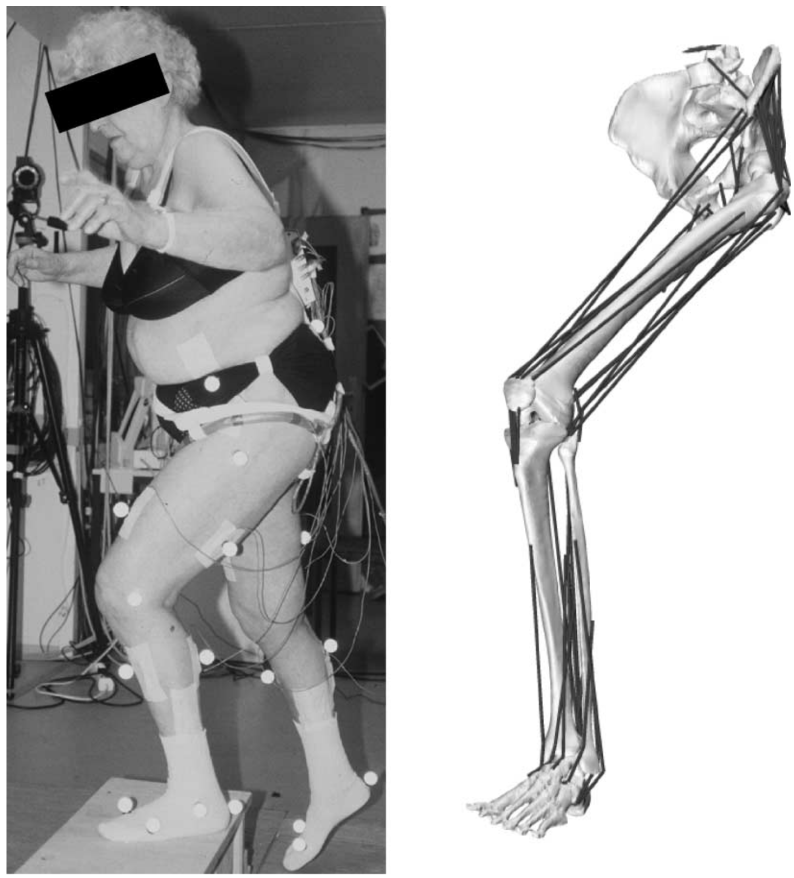
Four total hip arthroplasty (THA) patients were included in the study. In all patients, an instrumented femoral prosthesis was used to measure the in vivo hip contact forces (Bergmann et al., 1988, 1993, 1995). All subjects gave informed consent to participate in the experiments and to the publication of their images and names. The study was approved by the local ethics committee of the Free University of Berlin. In two patients the prosthesis was implanted in the left hip, in the others in the right hip. At the time of surgery, the mean age of the patients was 61 years, ranging from 51 to 76 years. The mean time between surgery and measurements was 17 months, ranging from 11 to 31 months. For each patient, extensive anthropometric data was collected to determine bone dimensions, segment masses, center of gravity positions and inertia parameters (Table 1) (Deuretzbacher and Rehder, 1995).

2.2. Gait analysis and inverse dynamics

Clinical gait analysis was conducted for a number of different activities of every day life (Bergmann et al., 2001). The present study concentrates on activities with the highest frequencies during daily living (Morlock et al., 2001), such as walking and stair climbing. Each patient performed several trials of each activity. The average speed during walking was 3.9 km/h. The stair climbing exercise was performed on custom made stairs composed of three single steps (step height: 17 cm) without hand rail support. A stair depth of 45 cm was determined from measuremens of the patients average stride length. Three patients climbed all steps (HSR, PFL, KWR) while patient IBL climbed only the first one. All measurements were taken during climbing up the first step. Start and end of the walking and stair climbing exercises were determined by instants of heel contact. The beginning of the exercise was defined as the moment of heel strike (0% stride). The end of the exercise was marked by the next heel strike of the same leg (100% stride). During all activities, time dependent kinematic and kinetic data were gathered: two Kistler force plates measured ground reaction forces. The in vivo hip contact force with the magnitude F and the components $-F_x$, $-F_y$, $-F_z$ was measured in the femur system (x, y, z) during all activities. The x axis of the femur system is parallel to the dorsal contour of the femoral condyles in the transverse plane, the z axis is parallel to an idealized midline of the femur (Bergmann et al., 1993). An optical system (Vicon, Oxford Metrics, UK) consisting of a set of six infrared cameras and 24 reflective markers attached to the patients' skin was used to determine movement of the lower limbs (Fig. 1a).

Positional information given by the skin markers and the anthropometric data (Table 1) were combined to derive the locations of bony landmarks (Table 2). From these landmarks, the segment coordinate systems (origins and orientations) were computed with respect to the fixed gait laboratory system. The resultant intersegmental forces and moments at ankle, knee and hip joint were computed from the kinematic and kinetic gait analysis data with respect to the local coordinate systems using an inverse dynamics approach (Andrews, 1974; Deuretzbacher and Rehder, 1995).

Peak values of the vertical force during a gait cycle were used to define inter-individual variability. The mean vertical peak force was computed as the arithmetic mean of the peak vertical forces of all trials performed



(a) (b)

Fig. 1. (a) The photograph shows patient IBL climbing stairs. The markers attached to the skin and a set of six infrared cameras (only one camera is visible in the background of the picture) were used to obtain the kinematic data. The coil visible around the patient's left hip was used to power the electronics for measurement of the in vivo hip contact forces. (b) The musculo-skeletal model of this patient when going upstairs.

Table 1
Anthropometric data used to develop individual patient musculo-skeletal models. Lengths were defined by the distance between two bony landmarks (see Table 2): pelvic width: landmarks 2 and 3, thigh length: landmarks 4 and 5, shank length: landmarks 6 and 7, foot length: landmarks 8 and 9

Patient	IBL	HSR	PFL	KWR
Instrumented side	Left	Right	Left	Right
Age at surgery (years)	76	55	51	61
Time after surgery (months)	31	14	11	12
Anteversion angle (°)	14	4	23	−2
Lengths (cm)				
Body height	170.0	154.0	175.0	165.0
Pelvic width	18.7	16.5	14.5	14.6
Thigh length	42.3	39.3	37.0	34.9
Shank length	40.5	38.2	40.7	40.6
Foot length	21.0	23.5	25.8	24.7
Weights (N)				
Body weight	800	878	972	703
Weight thigh	98.0	76.2	80.5	62.2
Weight shank	43.3	38.9	53.7	36.6
Weight foot	7.2	9.1	12.2	9.7

Table 2
Description of the bony landmarks used to define the local segment coordinate systems of pelvis, thigh, shank and foot

Landmark	Segment	Description
1	Pelvis (origin)	L5-S1 transition
2	Pelvis	Right hip joint center
3	Pelvis/left thigh (origin)	Left hip joint center
4	Left thigh	Neck-shaft transition prosthesis
5	Left thigh/left shank (origin)	Left knee joint center
6	Left shank	Left lateral articular space
7	Left shank/left foot (origin)	Left ankle joint center
8	Left foot	Left calcaneal tuberosity
9	Left foot	Left proximal phalanx of 5th digit

by a single patient. The relative variability over all trials in percent was defined as the maximum difference between the peak force of a single trial and the mean peak force.

Similarly, inter-individual variability of the flexion–extension moment at the hip was defined. The mean peak moment was computed as the arithmetic mean of the peak flexion-extension moments of all trials performed by a single patient. Relative variability over all trials in percent was defined as the maximum difference between the peak moment of a single trial and the mean peak moment.

To allow inter-individual comparison, the data was mirrored for those patients with a prosthesis on the right side. Thus, all data were available for a prosthesis on the left side.

2.3. Musculo-skeletal model

Based on CT-data from the Visible Human (NLM, Bethesda, USA), a musculo-skeletal model of the human lower extremity was developed. The Visible Human data set was chosen as it is the most complete, official dataset available that describes the human anatomy. The CT-scans were available at a spacing of 1 mm with a slice thickness of 1 mm (in plane scan resolution: 0.9375 mm/pixel) and thus allowed an accurate description of bony anatomy to be obtained. From the CT-scans, surface contour data of all hip bones (left and right iliac bone, sacrum and vertebra S1) and all the bones of the left leg (femur, patella, tibia, fibula, and all the bones of the foot) were determined. This was achieved by thresholding methods available in the "Medical Image Editor" software by courtesy of the German Heart Center, Berlin. The bony surfaces were reconstructed from the contours (Geiger, 1993). To remove noise, the surfaces were filtered by a polygonal mesh filter available in the Visualization Tool Kit (Schroeder et al., 1998).

Muscles were represented as straight lines spanning from origin to insertion based on descriptions from the literature (Brand et al., 1982; Duda et al., 1996a). Muscles with large attachment areas such as the glutei were modeled by more than one line of action. Some muscles were wrapped around the bones to approximate their real curved path. This was necessary to gain an adequate representation of their lever arms at the joints. In total, the muscle model included 95 lines of action. Data on the physiological cross sectional area (PCSA) of the individual muscles was taken from the literature (Brand et al., 1986; Duda et al., 1996a).

2.4. Joint kinematics

Hip and ankle were modeled as joints with three rotational degrees of freedom (DOF). At the knee, the femoro-tibial joint was modeled as a joint with three rotational DOF while the patello-femoral joint was modeled as a joint with one rotational DOF around the medio-lateral axis, and two translational DOF in the sagittal plane.

The tracking of the patella during gait analysis was impossible. An in vitro experiment was conducted to determine the kinematics of the patello-femoral joint. A human knee specimen was mounted in a knee joint simulator allowing unconstrained knee motion and loading (Dürselen et al., 1995). The motion of the patella in the sagittal plane was derived for a complete flexion-extension cycle of the knee. In order to adapt the data obtained from the knee specimen to the patients, anterior–posterior and axial translations were scaled based on the patella position in full extension of the knee.

2.5. Adaptation to the individual patient

Adaptation of the Visible Human to the individual patient anatomy was accomplished by a scaling process. The procedure employed bony landmarks between which bone dimensions were defined. These landmarks were determined for the Visible Human and the patients (Table 2). Scaling factors were calculated as the ratio of patients' to Visible Human bone dimensions. Linear scaling was applied individually to each bone and all attached muscle origins, insertions or wrapping points in order to obtain individual patient musculo-skeletal models.

Medio-lateral scaling of the pelvis was based on the distance between the left and right hip joint centers. The thigh was scaled to match the length between the transition point of prosthesis neck and shaft and the knee joint center. The same scaling factor was applied to the patella. The distance between knee and ankle joint center was used to compute the scaling parameter for the shank. The foot was scaled based on the distance between calcaneal tuberosity and the phalanx of the fifth digit. The PCSA of each muscle was scaled according to the patients' body weight.

Head and neck of the Visible Human femur were resected to simulate the THA surgery. In a first step, the proximal part of the femoral prosthesis (neck modeled as a cylinder, prosthesis head modeled as a sphere) was scaled based on the patient's neck length and head diameter. In a second step, neck and head were positioned and oriented towards the resected femur according to femoral anteversion, caput collum, diaphyseal and neck-stem angle to match the individual implantation.

2.6. Computation of muscle and joint contact forces

Muscle force distribution was computed with a linear optimization algorithm, minimizing the sum of muscle forces (Crowninshield, 1978). Inequality constraints were imposed on the maximum muscle forces (Challis, 1997). Maximum muscle activation during the every day activities under investigation was unlikely to occur.

Therefore, the muscle forces were restricted to below 85% of a physiological muscle force. This force was calculated as the product of each muscle's PCSA and a physiological muscle stress of 1.0 MPa (An et al., 1989).

A distribution of muscle forces was required to fulfill the resultant intersegmental moments at ankle (flexion-extension moment), knee (flexion-extension and ab-adduction moments) and hip joint (all moments). From the individual muscle and the resultant intersegmental forces, joint contact forces were calculated for ankle, knee and hip joints for all patients and activities. The calculated hip contact forces and those measured in vivo were compared for all trials.

3. Results

3.1. Ground reaction forces

The general pattern and the magnitudes of the ground reaction forces were similar for all trials involving one individual patient during walking (Fig. 2a). The ground reaction forces were characterized by a dominant, vertically directed component. The relative variability of the vertical peak forces for a single patient ranged from 2% to 5% with an average of 4% for all patients (IBL: 2%, HSR: 4%, PFL: 5%, KWR: 5%). Findings for stair climbing were similar (Fig. 2b). Again, vertical force dominated. Relative variability of vertical peak forces for a single patient ranged from 1% to 6% with an average of 4% for all patients (IBL: 5%, HSR: 4%, PFL: 1%, KWR: 6%).

3.2. Resultant intersegmental moments at the hip

For both walking and stair climbing, the general characteristics of the resultant intersegmental moments at the hip were similar for the different trials for each patient (Fig. 3a, b). The largest moment was always the flexion–extension moment. The average relative variability in the flexion extension moment for all patients was 19% during walking (IBL: 11%, HSR: 27%, PFL: 27%, KWR: 11%) and 11% during stair climbing (IBL: 21%, HSR: 5%, PFL: 4%, KWR: 15%).

3.3. Measured vs. calculated hip contact forces

Calculated hip contact forces and those measured in vivo were compared in all investigated trials. Graphical representation is restricted to one trial per patient (Fig. 4a, b).

The cycle-to-cycle comparison revealed good agreement in pattern and magnitudes of computed and measured hip contact forces for walking in all four patients. Relative deviation was defined as the difference between measured and calculated hip contact forces divided by the measured force and evaluated for each moment during the gait cycle. During the stance phase where absolute forces were much larger than during the swing phase, relative deviations in absolute hip contact force magnitudes were smallest (Fig. 4a). The smallest relative deviation of all three force components was found for the axially directed component F_z. The smaller, medio-lateral and anterior–posterior directed contact forces F_x and F_y showed larger relative deviations, both under or over-estimating the in vivo measured forces. At the moment of maximum measured hip contact force, the minimal relative deviation between measured and calculated hip contact forces for all trials was 0.3% (patient KWR). In the trial with the largest deviation, the calculation overestimated the hip contact force by 33% (patient HSR). The arithmetic mean of the relative deviation of absolute measured and calculated force magnitudes during walking was 12% for all patients (mean values determined from all trials for the individual patients: IBL: 13%, HSR: 23%, PFL: 10%, KWR: 2%).

The findings for stair climbing were similar. General pattern and magnitudes of the calculated hip contact forces agreed well with the in vivo measured data (Fig. 4b), especially during the stance phases. The smallest relative deviation was found for the axially directed contact force component F_z, while the medio-lateral and anterior–posterior forces F_x and F_y showed larger relative deviations. For a single trial of an individual patient, the smallest relative deviation between measured and calculated absolute force magnitudes at the moment of maximum measured hip contact force was 3%, the largest 37%. Relative deviation of the absolute force magnitudes during stair climbing showed a mean of 14% for all patients (mean values determined from all trials for the individual patients: IBL: 8%, HSR: 15%, PFL: 21%, KWR: 13%).

Overall, there seems to be a tendency for an overestimation of the hip contact forces by the optimization approach (Fig. 5).

4. Discussion

The aim of this study was to determine the musculoskeletal loading conditions of the proximal femur during walking and stair climbing. While previous studies computed muscle forces (Brand et al., 1986; Collins, 1995; Davy and Audu, 1987; Fuller and Winters, 1993; Glitsch and Baumann, 1997; Herzog, 1987; Pedersen et al., 1987; Pedersen et al., 1997; Röhrle et al., 1984; Seireg and Arvikar, 1973, 1975; Siebertz and Baumann, 1994), a direct validation of the predicted loading conditions at the hip was not possible. The study of Brand et al. (1994) presented measured and calculated hip contact forces in the same patient, but a

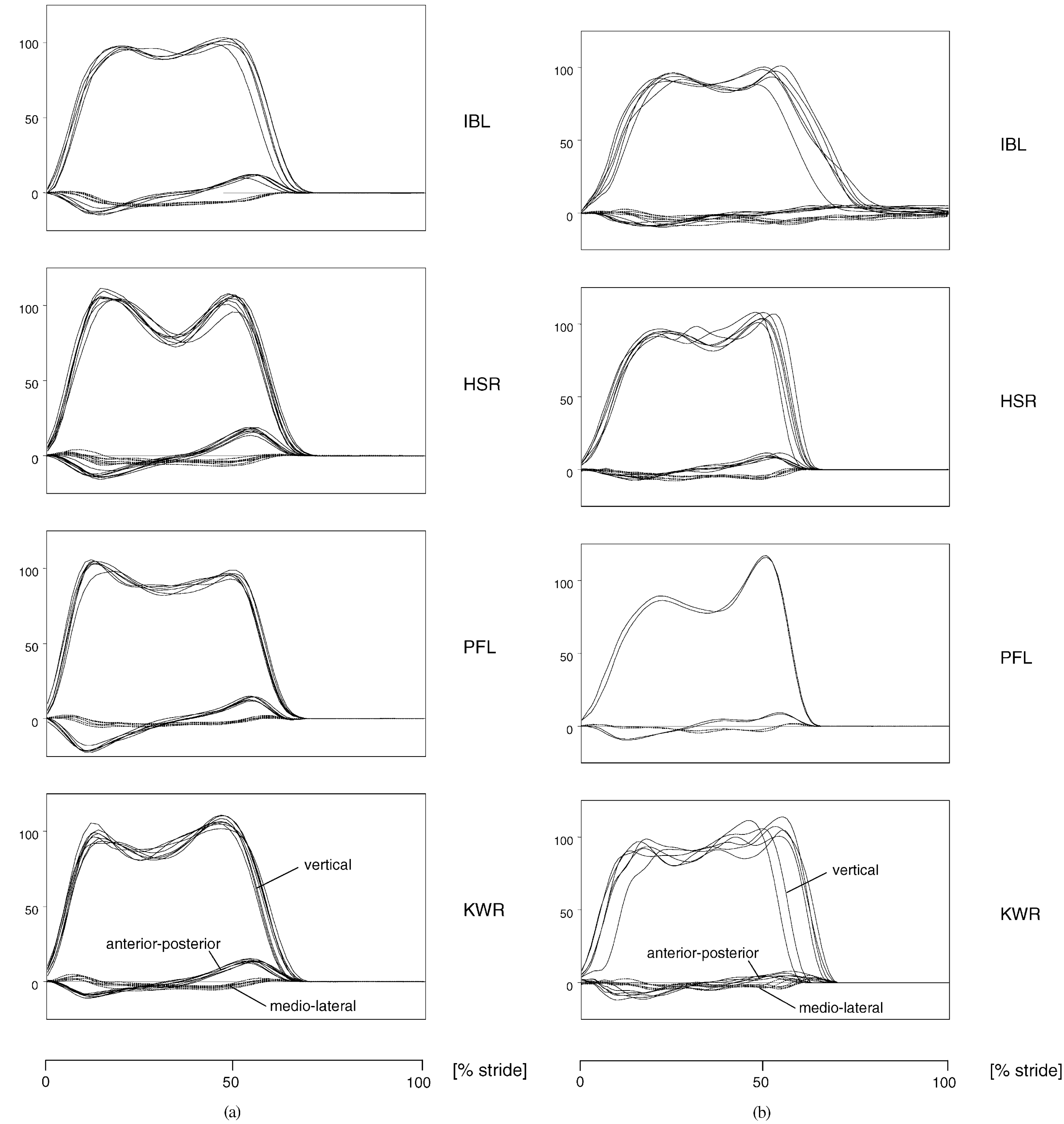


Fig. 2. Three components of the ground reaction force acting at the foot for all trials of four patients during; (a) walking and (b) stair climbing. Values are given in percent of body weight. Due to data errors, only two trials of stair climbing were evaluated for patient PFL.

cycle-to-cycle comparison was not possible The calculated peak forces seemed to be somewhat larger than the measured forces, but a quantification of the differences was not possible.

In the present study, a musculo-skeletal model of the human lower extremity was established and validated by a cycle-to-cycle comparison of in vivo measured and calculated hip contact forces. The quantitative comparison supports the general tendency for an overestimation of the hip forces by the calculation found by Brand et al. (1994).

Four THA patients with a telemetering prosthesis to measure the in vivo hip contact forces were included in this study. Since the participants of this study underwent THA, it may not be expected that the loading conditions in these patients are identical to those in healthy

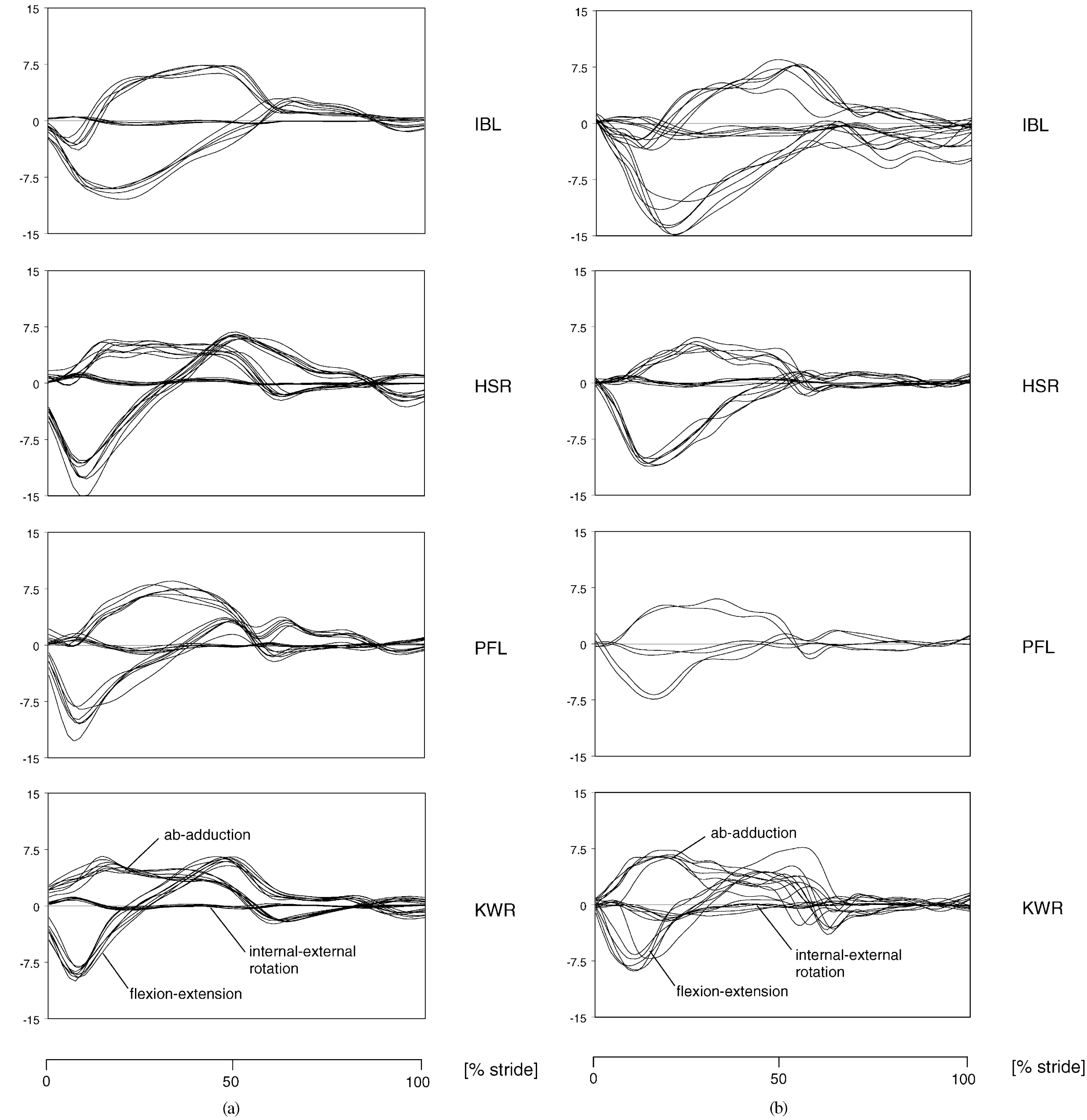


Fig. 3. Three components of the intersegmental moment at the hip during; (a) walking and (b) stair climbing, for all trials of four patients. Moments are given in percent body weight times metre. Positive moments cause abduction, internal rotation and flexion, respectively.

subjects. Nevertheless, the approach presented here represents the only methodology to achieve a validation with respect to hip contact forces.

Ground reaction forces were used to indicate the reproduciblity of the gait pattern in different trials of the same activity. Large deviations in the pattern and magnitudes of the ground reaction forces would have revealed variations in the patients' performance of activities, e.g. in different trials of walking. The small variations observed for the ground reaction forces at the same speed suggest that the patients walked in a quite similar and reproducible way.

It is well known that skin movement errors can affect location of bony landmarks derived from the marker positions. Therefore, special care was taken to minimize skin movements and other artifacts. It turned out that the longitudinal scaling of the bones was virtually unaffected by errors in the spatial position of the landmarks. However, sometimes the internal–external rotation of the thigh revealed erroneous axial rotation

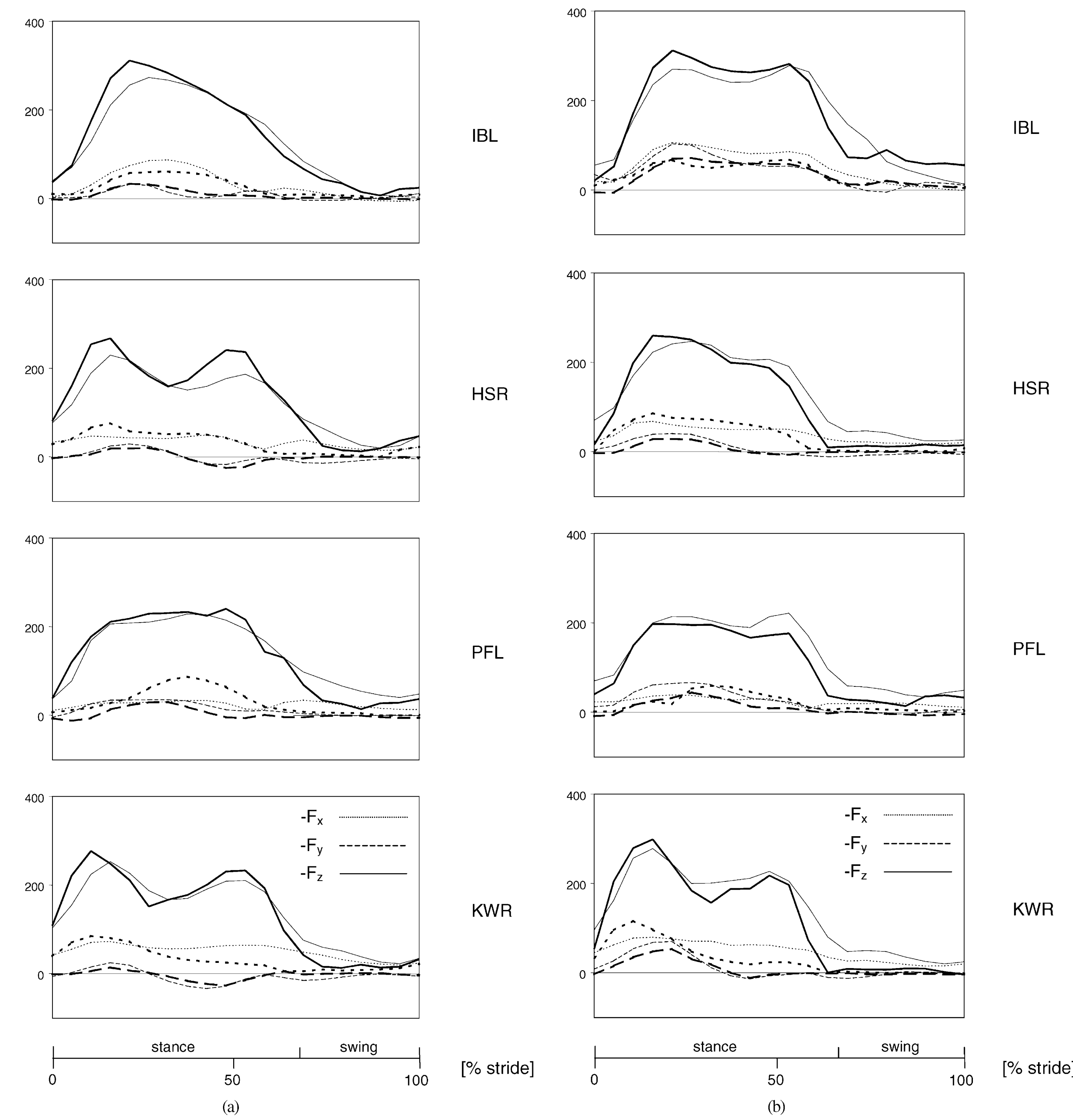


Fig. 4. Measured (thin lines) and calculated (thick lines) three-dimensional hip contact forces of four patients during; (a) walking and (b) stair climbing. Values in percent of body weight. The force components $-F_x$, $-F_y$, $-F_z$ are given in the 'femur system' x, y, z. The x axis of the femur system is parallel to the dorsal contour of the femoral condyles in the transverse plane, the z axis is parallel to an idealized midline of the femur (Bergmann et al., 1993).

data, as the femoral condyles did not match the tibia. To address this problem, the 3-D model allowed correction of the rotation of the affected limb by controlling the congruency of the femoral condyles and the tibial plateau. The method recently proposed by Lu and O'Connor (1999) might help to avoid or at least substantially reduce detrimental effects of skin movement in future studies.

Because the inverse dynamics calculation is an iterative process starting from the ankle joint, the largest errors due to error propagation and error accumulation were most likely to occur at the most proximal joint in the model, the hip joint. While the ground reaction forces showed an intra-individual variation of 4%, the flexion-extension moments at the hip varied by as much as 19%. The impact of error propagation or actual

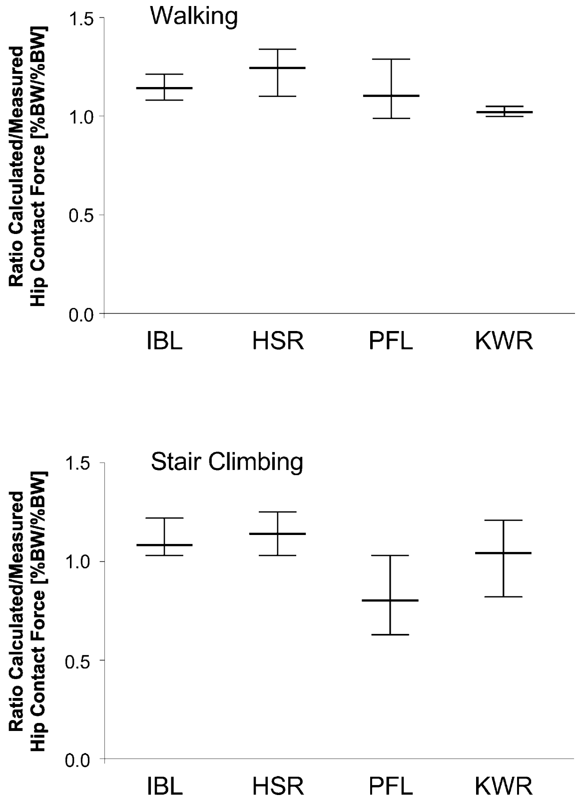


Fig. 5. Ratio of calculated to measured hip contact forces at the instance of maximum measured hip contact force. A ratio larger than 1.0 means an overestimation of the hip contact force by the optimization approach. Similarly, a ratio smaller 1.0 indicates underestimation of the hip contact force. The diagramms show the mean values as well as the range for all trials of walking and stair climbing.

intra-individual variations on the observed findings remains to be clarified.

In order to predict musculo-skeletal loading conditions, an accurate model of bones and muscles was mandatory. An accurate model of human anatomy was developed based on the Visible Human data set. The narrow spacing of the CT-scans and the high in plane resolution of the CT-scans allowed a good description of the bony anatomy and muscle attachments. Muscles were modeled as straight lines. Wrapping of muscles was used to simulate their real force distribution and lever arms relative to the joints. Nevertheless, the actual three-dimensional volumetric structures and curved pathways of the muscles had to be simplified. This might explain why the correlation between hip contact forces measured in vivo and calculated hip contact force components in the transverse plane was not as good as for the axial component (Jensen and Davy, 1975). To overcome this limitation, current work is focusing on a more complex modeling of the 3-D volumetric structure of the muscles (Pohl et al., 1999).

Patello-femoral kinematics based on data obtained from an in vitro study were included in order to approximate the lever arms of the muscles attached to the patella, especially for the larger knee flexion angles which occurred during stair climbing. This behavior of the model might be improved further to investigate femoro-tibial translation as well. However, the implementation of the full complexity of knee kinematics as proposed by other authors (e.g. Lu et al., 1998; Wilson et al., 1998) was not within the scope of this study. During walking, the flexion angle was only moderate. While the flexion angles were greater during stair climbing, good agreement between calculated and in vivo measured hip contact forces was found when the patello-femoral kinematics were included. Future studies will show whether further improvements in the representation of joint kinematics will also allow more extensive application of the model to include other activities such as knee bends.

The Visible Human anatomy was scaled to each patient to permit calculations based on individual musculo-skeletal models. With the information obtained from routine CT-scans and X-ray data, important features such as femoral anteversion and bone dimensions were determined. Important anatomical features such as pelvic width, femoral, tibial and foot length were approximated using the scaling technique presented here. A more thorough scaling technique may further improve the results of the cycle-to-cycle comparisons.

The optimization approach used to estimate muscle forces was similar to that previously used in other studies. Consequently, all limitations discussed therein also apply to the present study, e.g. the dependency of individual muscle forces on PCSA (Brand et al., 1986) or the dependency of individual muscle forces on the objective function employed in the optimization calculation (Pedersen et al., 1987). The linear objective function employed in this study might not be capable of predicting complex muscle activation (Crowninshield and Brand, 1981) and the individual muscle forces might be high. To avoid this, the forces were constrained to stay below critical threshold levels (An et al., 1989). The results of this study indicate that linear optimization can be used to predict proximal femoral loading which is comparable to in vivo data.

5. Conclusions

The musculo-skeletal model of the lower extremity presented in this study allowed prediction of proximal femoral loading for walking and stair climbing in four THA patients. Although the patients were of different ages and the implantation varied considerably in the anteversion angle, the musculo-skeletal loading conditions were characterized by similar patterns and magnitudes.

With walking and stair climbing two daily activities associated with both a large number of load cycles and large hip contact forces were investigated (Morlock et al., 2001). The calculated hip contact forces and those measured in vivo during these every day activities were similar. However, a varying degree of conformity between the individual force components was found. The component acting along an idealized femoral midline showed best agreement, while the results for the significantly smaller forces in the transverse plane were less accurate. For the first time, a direct cycle-to-cycle validation of proximal femoral loading was possible. The cycle-to-cycle validation revealed that absolute peak loads differed by an average of only 12% during walking and 14% during stair climbing.

In order to predict musculo-skeletal loading conditions, two issues seem to be important. First, a suitable measuring procedure to validate the prediction should be accessible. In this study, the in vivo measured hip contact forces can be used for cycle-to-cycle validation of the predicted hip contact forces. Second, patient individual models should be used to approximate the loading conditions in each individual case. The biomechanical model used in the present study was adapted to the individual anatomy and prosthesis configuration.

In summary, the authors consider the model presented here to be a means to determine mechanical boundary conditions for implant loading, bone remodeling and fracture healing. It opens up the prospect of investigating the implant–bone load sharing and primary stability of implants under loading conditions which approximate most realistically to the in vivo loading conditions of walking and stair climbing. In addition, analysis of exercises usually carried out during rehabilitation can be of help in identifying activities with possibly hazardous musculo-skeletal loading and thus improve the outcome of rehabilitation.

Acknowledgements

This study was supported by the EC (contract SMT-CT96-207).

References

An, K.N., Kaufman, K.R., Chao, E.Y.S., 1989. Physiological considerations of muscle force through the elbow joint. Journal of Biomechanics 22, 1249–1256.

Andrews, J.G., 1974. Biomechanical analysis of human motion. Kinesiology 4, 32–42.

Bergmann, G., Deuretzbacher, G., Heller, M.O., Graichen, F., Rohlmann, A., Strauss, M., Duda, G., 2001. Hip contact forces and gait patterns from routine activities. Journal of Biomechanics 34, 859–871.

Bergmann, G., Graichen, F., Rohlmann, A., 1993. Hip joint loading during walking and running, measured in two patients. Journal of Biomechanics 26, 969–990.

Bergmann, G., Graichen, F., Rohlmann, A., 1995. Is staircase walking a risk for the fixation of hip implants? Journal of Biomechanics 28, 535–553.

Bergmann, G., Graichen, F., Siraky, J., Jendrzynski, H., Rohlmann, A., 1988. Multichannel strain gauge telemetry for orthopaedic implants. Journal of Biomechanics 21, 169–176.

Brand, R.A., Crowninshield, R.D., Wittstock, C.E., Pedersen, D.R., Clark, C.R., van Krieken, F.M., 1982. A model of lower extremity muscular anatomy. Journal of Biomechanic Engineering 104, 304–310.

Brand, R.A., Pedersen, D.R., Davy, D.T., Heiple, K.G., Goldberg, V.M., 1989. Comparison of hip force calculations and measurements in the same patient. Transactions of ORS 1, 96–99.

Brand, R.A., Pedersen, D.R., Davy, D.T., Kotzar, G.M., Kingsbury, G.H., Goldberg, V.M., 1994. Comparison of hip force calculations and measurements in the same patient. Journal of Arthroplasty 9, 45–51.

Brand, R.A., Pedersen, D.R., Friederich, J.A., 1986. The sensitivity of muscle force predictions to changes in physiological cross-sectional area. Journal of Biomechanics 19, 589–596.

Challis, J.H., 1997. Producing physiologically realistic individual muscle force estimations by imposing constraints when using opimization. Medical Engineering Physics 19, 253–261.

Cheal, E.J., Spector, M., Hayes, W.C., 1992. Role of loads and prosthesis material properties on the mechanics of the proximal femur after total hip arthroplasty. Journal of Orthopaedic Research 10, 405–422.

Claes, L.E., Heigele, C.A., Neidlinger-Wilke, C., Kaspar, D., Seidl, W., Margevicius, K.J. and Augat, P., 1998. Effects of mechanical factors on the fracture healing process. Clinical Orthopaedics 355 Suppl S132–147.

Collins, J.J., 1995. The redundant nature of locomotor optimization laws. Journal of Biomechanics 28, 251–267.

Crowninshield, R.D., 1978. Use of optimization techniques to predict muscle forces. Journal of Biomechanical Engineering 100, 88–92.

Crowninshield, R.D., Brand, R.A., 1981. A physiologically based criterion of muscle force prediction in locomotion. Journal of Biomechanics 14, 793–801.

Davy, D.T., Audu, M.L., 1987. A dynamic optimization technique for predicting muscle forces in the swing phase of gait. Journal of Biomechanics 20, 187–201.

Davy, D.T., Kotzar, G.M., Brown, R.H., Goldberg, V.M., Heiple, K.G., Berilla, J., Burstein, A.H., 1988. Telemetric force measurement across the hip after total hip arthroplasty. Journal of Bone and Joint Surgery (Am.) 70-A, 45–50.

Deuretzbacher, G., Rehder, U., 1995. A CAE (computer aided engineering) approach to dynamic whole body modeling—the forces in the lumbar spine in asymmetrical lifting. Biomediziniscte Technik (Berlin) 40, 93–98.

Duda, G. N., 1996. Influence of muscle forces on the internal loading in the femur during gait. Technische Universität Hamburg-Harburg.

Duda, G.N., Heller, M., Albinger, J., Schulz, O., Schneider, E., Claes, L., 1998. Influence of muscle forces on femoral strain distribution. Journal of Biomechanics 31, 841–846.

Duda, G.N., Brand, D., Freitag, S., Lierse, W., Schneider, E., 1996a. Variability of femoral muscle attachments. Journal of Biomechanics 29, 1183–1190.

Dürselen, L., Claes, L., Kiefer, H., 1995. The influence of muscle forces and external loads on cruciate ligament strain. American Journal of Sports and Medicine 23, 129–136.

English, T.A., Kilvington, M., 1979. In vivo records of hip loads using a femoral implant with telemetric output (a preliminary report). Journal of Biomedical Engineering 1, 111–115.

Frost, H.M., 1999. Why do bone strength and "mass" in aging adults become unresponsive to vigorous exercise? Insights of the Utah paradigm. Journal of Bone Mineral Metabolism 17, 90–97.

Fuller, J.J., Winters, J.M., 1993. Assessment of 3-D joint contact load predictions during postural/stretching exercises in aged females. Annals of Biomedical Engineering 21, 277–288.

Geiger, B., 1993. Three dimensional modeling of human organs and its application to diagnosis and surgical planning. Institut National de Recherche en Informatique et Automatique. Technical report 2105.

Glitsch, U., Baumann, W., 1997. The three-dimensional determination of internal loads in the lower extremity. Journal of Biomechanics 30, 1123–1131.

Herzog, W., 1987. Individual muscle force estimations using a non-linear optimal design. Journal of Neuroscience Methods 21, 167–179.

Jensen, R.H., Davy, D.T., 1975. An investigation of muscle lines of action about the hip: a centroid line approach vs the straight line approach. Journal of Biomechanics 8, 103–110.

Lu, T.-W., O'Connor, J.J., 1999. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. Journal of Biomechanics 32, 129–134.

Lu, T.-W., O'Connor, J.J., Taylor, S.J.G., Walker, P.S., 1998. Validation of a lower limb model with in vivo femoral forces telemetered from two subjects. Journal of Biomechanics 31, 63–69.

Morlock, M., Schneider, E., Bluhm, A., Vollmer, M., Bergmann, G., Müller, V., Honl, M., 2001. Duration and frequency of every day activities in total hip patient. Journal of Biomechanics 34, 873–888.

Pedersen, D.R., Brand, R.A., Cheng, C., Arora, J.S., 1987. Direct comparison of muscle force predictions using linear and nonlinear programming. Journal of Biomechanical Engineering 109, 192–199.

Pedersen, D.R., Brand, R.A., Davy, D.T., 1997. Pelvic muscle and acetabular contact forces during gait. Journal of Biomechanics 30, 959–965.

Pohl, M., Duda, G.N., Heller, M., Claes, L., 1999. Ein muskuloskelettales Modell der hinteren Schafsextremität zur Opimierung und Einschränkung tierexperimenteller Studien. Hefte zur Zeitschrift "Der Unfallchirurg" 275, 272–273.

Rohlmann, A., Mossner, U., Bergmann, G., Hees, G., Kolbel, R., 1983. [Stresses on the femur following hip joint replacement]. Zertschrift fuer Orthopaedic 121, 47–57.

Röhrle, H., Scholten, R., Sigolotto, C., Sollbach, W., Kellner, H., 1984. Joint forces in the human pelvis-leg skeleton during walking. Journal of Biomechanics 17, 409–424.

Rydell, N.W., 1966. Forces acting in the femoral head-prosthesis. Acta Orthopaedica Scandinavica, Supplementum 88, 37–39.

Schroeder, W., Martin, K., Lorensen, B., 1998. The Visualization Toolkit, 2nd Edition. Prentice-Hall.

Seireg, A., Arvikar, R.J., 1973. A mathematical model for evaluation of forces in lower extremities of the musculo-skeletal system. Journal of Biomechanics 6, 313–326.

Seireg, A., Arvikar, R.J., 1975. The prediction of muscular load sharing and joint forces in the lower extremities during walking. Journal of Biomechanics 8, 89–102.

Siebertz, K., Baumann, W., 1994. Biomechanische Belastungsanalyse der unteren Extremität. Biomedizinische Technik (Berlin) 39, 216–221.

Wilson, D.R., Feikes, J.D., O'Connor, J.J., 1998. Ligaments and articular contact guide passive knee flexion. Journal of Biomechanics 31, 1127–1136.