Application and evaluation of biomechanical models and scores for the planning of total hip arthroplasty

Article *in* Proceedings of the Institution of Mechanical Engineers Part H Journal of Engineering in Medicine · December 2012

DOI: 10.1177/0954411912445261 · Source: PubMed

Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine

http://pih.sagepub.com/

Application and evaluation of biomechanical models and scores for the planning of total hip arthroplasty

Jörg Eschweiler, Lorenz Fieten, Jasmin Dell'Anna, Koroush Kabir, Sascha Gravius, Markus Tingart and Klaus Radermacher

Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine published online 10 May 2012

DOI: 10.1177/0954411912445261

The online version of this article can be found at:
http://pih.sagepub.com/content/early/2012/05/10/0954411912445261

Published by:

SAGE

http://www.sagepublications.com

On behalf of:

Institution of Mechanical Engineers

Institution of Mechanical Engineers

Special Issue Article

Institution of MECHANICAL ENGINEERS

Journal of ENGINEERING IN MEDICINE

Proc IMechE Part H:
J Engineering in Medicine
0(0) 1–13
© IMechE 2012
Reprints and permissions:
sagepub.co.uk/journalsPermissions.nav
DOI: 10.1177/0954411912445261
pih.sagepub.com
SAGE

Application and evaluation of biomechanical models and scores for the planning of total hip arthroplasty

Jörg Eschweiler[1,2], Lorenz Fieten[1], Jasmin Dell'Anna[1], Koroush Kabir[3], Sascha Gravius[3], Markus Tingart[2] and Klaus Radermacher[1]

Abstract

Intimate knowledge of the biomechanics of a given individual hip joint provides a potential advantage during the planning of total hip arthroplasty, and would thus have a positive influence over the outcome of such an intervention. In current clinical practise, the surgical planning is based solely on the status of the individual hip and its radiographic appearance. However, additional information could be gathered from the radiography to be used as input data for biomechanical models aimed at calculating the resultant force F_R within the hip joint.

An investigation of the biomechanical models by Pauwels, Debrunner and Iglič was performed, where the magnitude of F_R calculated by the models showed a favourable comparison to the *in-vivo* data from instrumented prostheses by Bergmann. The Blumentritt model returned abnormally high results. The computational results showed large variations for F_R orientation, which tends to depend more on the model used than on patient-specific parameters.

Furthermore, a discrepancy was found between the data gathered from instrumented prostheses and the *Standing Human Model* within the 'AnyBody Modeling System™' software by AnyBody Tech. Additionally, the variations in inter-rater and intra-rater errors made while localizing radiographic landmarks were analysed with respect to their influence on Babisch-Layher-Blumentritt (BLB)-scoring using the Blumentritt hip model.

Keywords

Biomechanical model, instrumented implant, total hip arthroplasty, hip joint, resultant force, landmark detection

Date received: 24 January 2012; accepted: 19 March 2012

Introduction

The hip joint is one of the largest weight-bearing structures in the human body. Abnormal and repetitive loading of the hip makes it susceptible to a variety of disorders, such as osteoarthritis.[1] Apart from genetic disposition, mechanical loading conditions have been accepted as the primary factor.[1] Quantifying the relevant biomechanical parameters of the hip joint is crucial for understanding this medical condition. Advanced familiarity of hip biomechanics is of great value to health care professionals involved in the field, and aids in understanding hip conditions and treatments options, such as total hip arthroplasty (THA).[2] THA is a widely used method to treat patients with degenerative hip osteoarthritis.[3] In 2010 more than 213,600 primary hip replacement and 36,500 revision hip replacement surgeries were performed in Germany.[4] It is recommended that the biomechanical conditions are considered during computer-assisted orthopedic surgery in order to achieve more desirable long-term results.[5,6]

Different biomechanical models can be implemented as clinical aids to evaluate the hip joint of the patient prior to, and post surgery. In current clinical practice the surgical planning is based solely on the status of the individual hip and its radiographic appearance. This radiographic information can additionally be used to

[1]Helmholtz-Institute for Biomedical Engineering, RWTH Aachen University, Germany
[2]Department for Orthopaedic and Trauma Surgery, University Hospital Aachen, Germany
[3]Department for Orthopaedic and Trauma Surgery, University Hospital Bonn, Germany

Corresponding author:
Jörg Eschweiler, Chair of Medical Engineering, Helmholtz-Institute for Biomedical Engineering, RWTH Aachen University, Pauwelsstraße 20, 52074 Aachen, Germany.
Email: eschweiler@hia.rwth-aachen.de

identify relevant anatomical landmarks as input data for biomechanical models. These models estimate the resultant hip joint force F_R as one of the major biomechanical parameters. However, the incorrect detection of anatomical landmarks may lead to erroneous results of the related simulation and analysis. Information derived from biomechanical *in-vivo* studies are needed and used to validate the underlying models, the development of surgical simulation, and the planning tools for THA. The simulation and planning tools are used to optimize the resulting hip joint loading, to prevent the progression of hip dysplasia, and to increase the longevity of the artificial joint. Personalized modelling, biomechanical simulation and load analysis of implants, and their individual boundary conditions have been advocated.[3]

The aim of this article is to compare different approaches towards modelling and analysis of the hip joint resultant force F_R using the biomechanical models described in the literature, and to point out the problems related to landmark identification based on a sensitivity analysis of these approaches related to incorrect landmark determination.

Hip joint biomechanics

Determination of the hip joint resultant force F_R – overview

Knowledge about F_R is of particular interest for a variety of reasons:

(a) the importance of understanding the possible relationships between hip mechanics and degenerative joint diseases;[7]
(b) the planning and design of surgical procedures and implants;[2]
(c) the simulation and evaluation of the potential outcome;
(d) obtaining clues about the pathogenesis of disease.[8]

Although direct measurement of F_R with instrumented prostheses has been demonstrated by Bergmann et al.,[9,10] this approach only provides post-operative information of joint forces acting on the implant, and it can only be used as a reference framework for implant design and basic research. In today's clinical routine, detailed patient-specific information of the *in-vivo* forces and their distribution is not available.[11] The only way for the surgeon to consider these forces in the framework of individual treatment planning is by the use of simplified models and estimates.

Different approaches can be used to estimate these forces. In biomechanics, inverse dynamics are traditionally understood as the process of computing forces and moments in anatomical joints based on motion analysis, in combination with measured ground reaction forces. This is based on a theoretical biomechanical model including more or less personalized kinematics and kinetics of the relevant anatomy.[2,3,12–16] Although motion data can be routinely acquired in a motion laboratory, the integration of this information into surgical planning, for example of THA, is not yet established within the clinical routine. The necessity of an individual adaptation of the underlying biomechanical model, based for example on segmented MRI data as a prerequisite of a valid inverse dynamic calculation of joint reaction forces, might be a hindering factor. At present, valid information for complex three-dimensional (3D) modelling is almost impossible to acquire for personalized surgery planning in a clinical routine. However, simplified biomechanical models based on two-dimensional (2D) or 3D image information have been proposed in literature and are partially used within the clinic, for example in reposition osteotomies or in THA.

The earliest biomechanical hip model was established by Friedrich Pauwels in 1935.[17–21] Using anterior–posterior (a.p.) X-ray images, information of the proximal femur, the pelvis, body weight (BW), and force of the gluteal abductor muscles (F_M) were used to construct an individualized biomechanical model and hence to calculate F_R. Although the related anatomical structures, such as the lever arm of the abductor muscles are reduced from 3D to 2D information, this approach provided a pragmatic and clinically feasible approach towards an estimation of F_R and for example the analysis of the influence of surgical measures changing the lever length of forces around the hip joint. Other groups also followed this basic approach or developed their own models.[22–37] Using such mathematical models, different intraoperative measures resulting in different joint kinematics can be simulated by varying single or multiple parameters, such as the neck–shaft relationship, while keeping all other parameters constant.[38] Furthermore, one could study many subjects at relatively low cost.[38]

Simplified biomechanical hip joint models

Model according to Pauwels. In Pauwels'[17] single-legged stance model (Figure 1), he assumed that F_R acting on the femoral head is created by the partial BW (F_{G5} = total BW minus the weight-bearing leg) and the force of the abductor muscles (F_M) inserting at the greater trochanter. F_R directly acts through the centre of rotation of the hip joint (HRC), which corresponds approximately to the centre of the femoral head. Pauwels pointed out that a normal hip equilibrium is achieved when BW is balanced by the force of the abductors. F_{G5} acting through its lever arm d_5 is in equilibrium with the force F_M acting through their lever arm d_M

$$F_{G_5} \cdot d_5 = F_M \cdot d_M \tag{1}$$

The direction of F_R is along a straight line that joins the centre of rotation of the femoral head and the intersecting

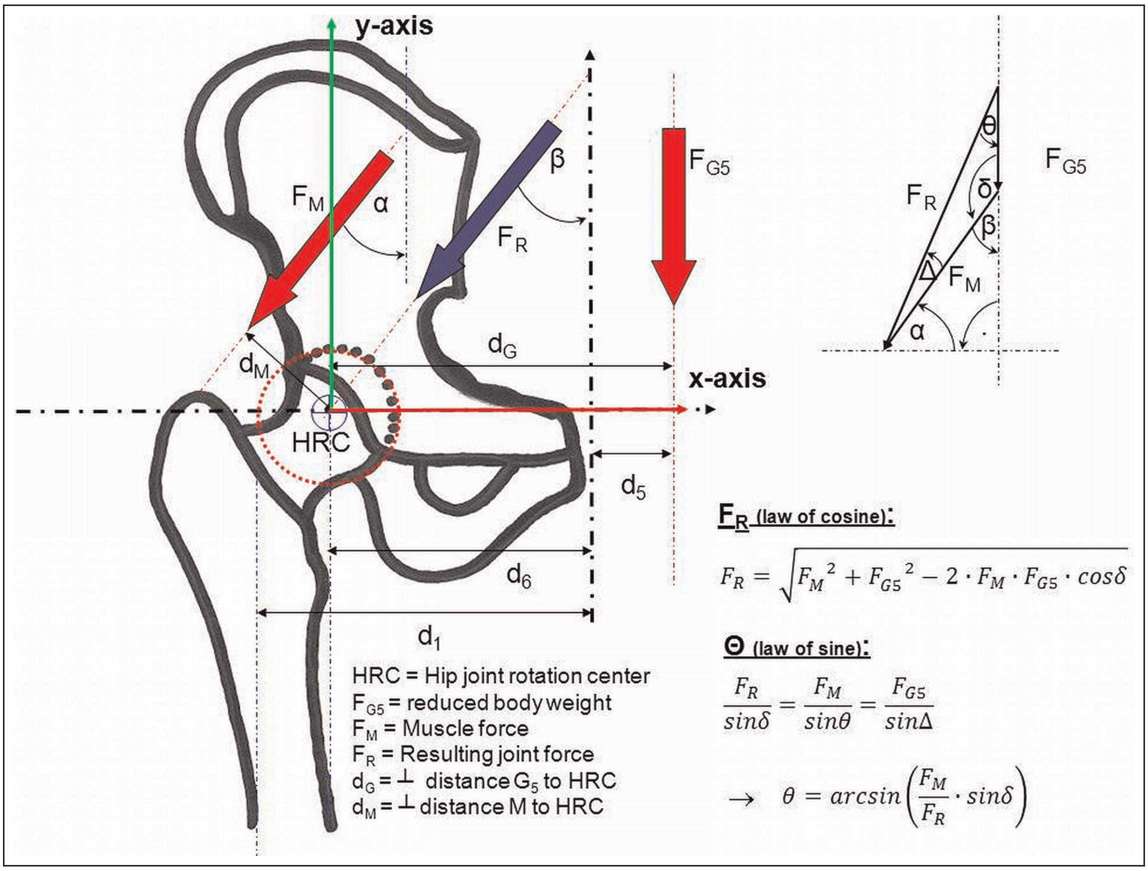


Figure 1. Simple mathematical model of the hip joint resultant force F_R and orientation (according to Pauwels[17]).

points of the lines of action of F_M and F_{G5} (Figure 1). Pauwels' model focuses on the single-legged stance.

However, questions about Pauwels' model arise with its simplicity and the lack of data on anatomical 3D structures. The magnitude of F_R basically depends on F_M, the partial BW F_{G5}, and the respective lever arm lengths. For a normal human body in a certain static single leg standing posture, the partial BW, acting through the lever arm d_5, should be definite. But, the abductor muscle force, acting through their lever arm d_M, is not fixed. Pauwels' model does not define a specific position for the line of action of the abductor muscle and is limited to 2D projections of a 3D anatomic situation.

Model according to Debrunner. Debrunner[23] developed a modified model to calculate the force F_R of the human hip joint. He adopted the basic idea of Pauwels' model. In his single-legged stance model, he also assumed that the resultant force F_R acting on the head of the supporting femur is created by the partial BW F_{G5_D} and the force of the abductor muscle F_{M_D}.

In contrast to Pauwels, Debrunner proposed a different approach towards modelling the origin point of the abductor muscle (see F_{M_D} in Figure 2) and the computation of the partial BW.[23] However, both approaches are very similar.

Model according to Blumentritt. Blumentritt analysed the mechanical loading of the hip joint based on a more detailed coxometric study, so that the generic biomechanical model could be adapted to the individual.[22,25] The calculation of the magnitude and direction of the hip joint resultant force F_R is based on the biomechanical model outlined in Figure 3.

The model is based on the angle and length measurements in a planar two-legged stance a.p. X-ray image of the hip. The equilibrium moment of the model is established by the pelvi-trochanteric muscle group (F_P), the spino-crural muscle group (F_{SP}), and the partial BW (F_{G5}), with an additional dynamic (acceleration) force (F_A) that is proportional to F_{G5}.

Note that this represents an inconsistency with natural gait. In a standing X-ray, pelvic orientation and position of the hip centre of rotation are different from the single-leg stance phase in a gait cycle (corresponding to maximal acceleration forces F_A). In the case of two-leg standing, F_A should be equal to zero. F_P is similar to the Pauwels' abductor muscles force F_M, and F_{SP} is the muscle force of the rectus femoris muscle. According to this biomechanical model, F_R can be calculated for its maximum load while walking with the following mathematical equations[22,25]

$$d_5(F_{G_5} + F_A) + lF_P \sin(\alpha - \beta) + sF_{SP} \sin(\Delta - \delta) = 0 \tag{2}$$

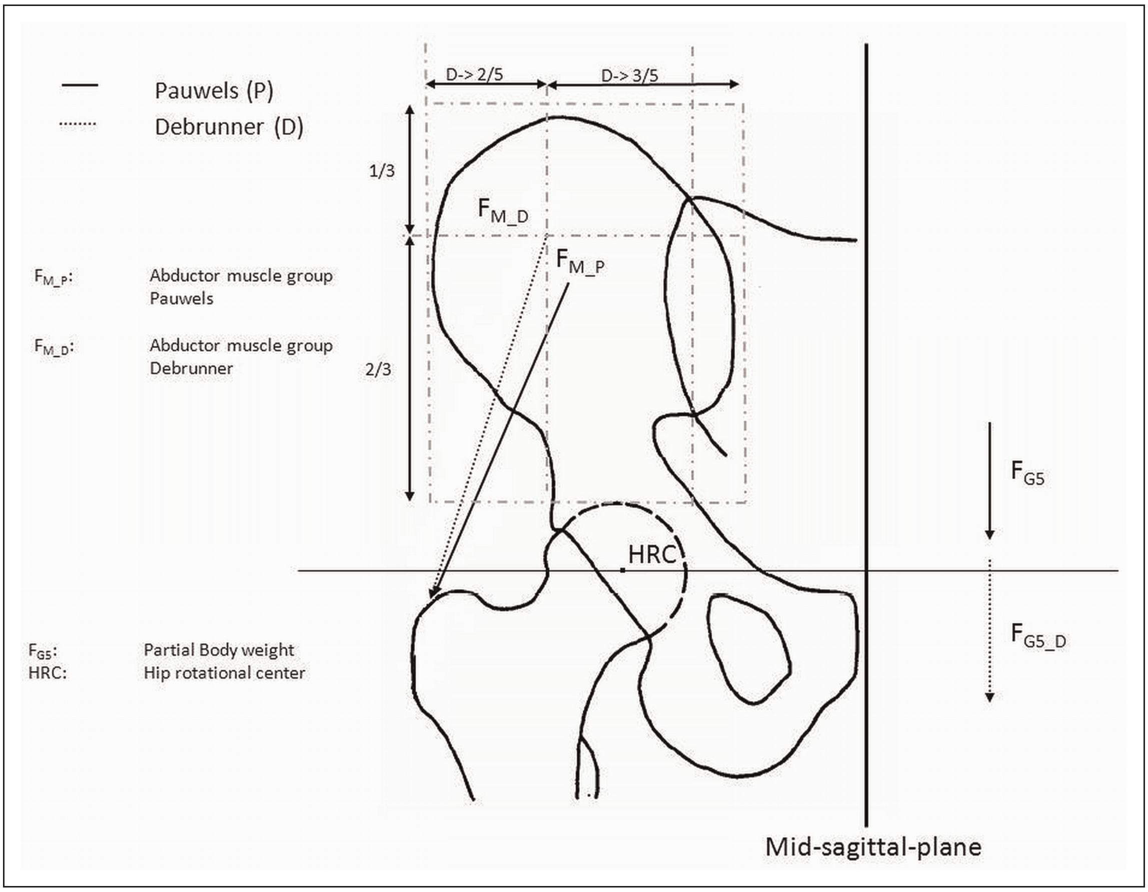


Figure 2. Differences in hip joint models (according to Pauwels,[17] Blumentritt,[22,25] and Debrunner[23].

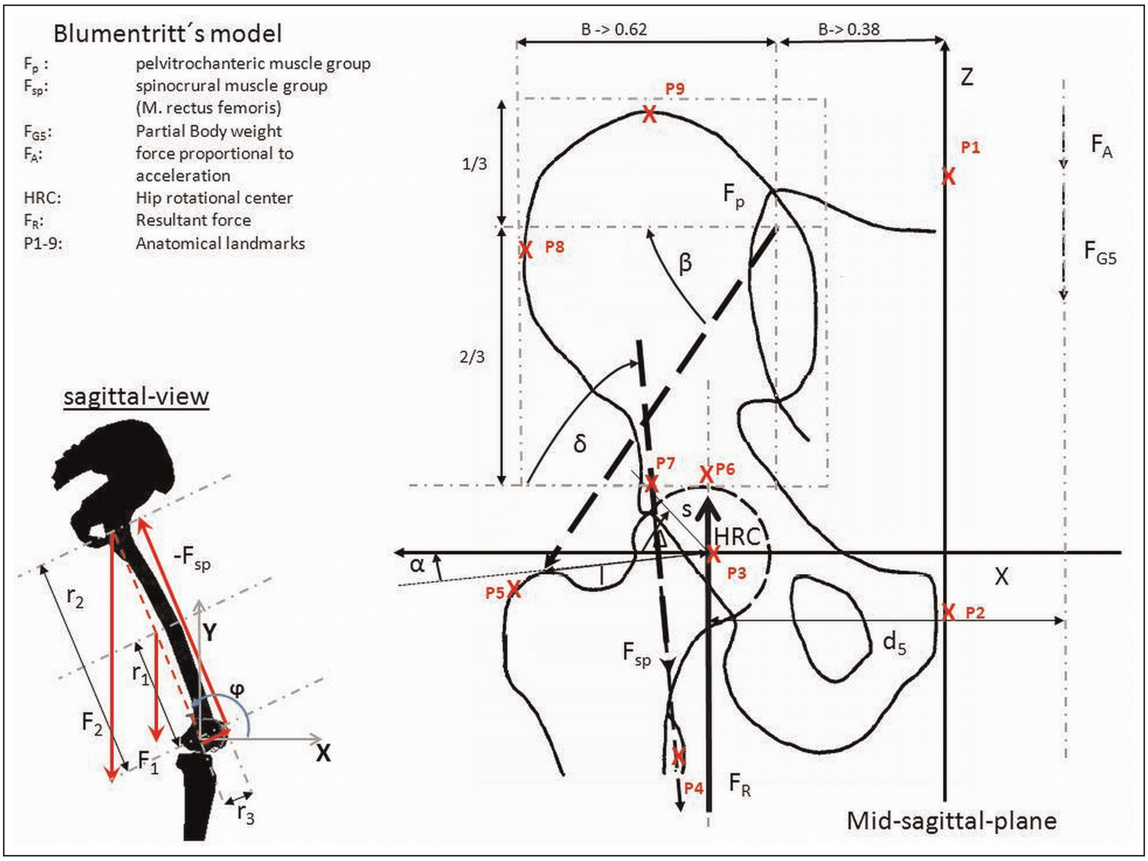


Figure 3. Blumentritt's model (according to Blumentritt[22,25]).

$$F_{SP}\cos(\delta) - F_P\cos(\beta) + F_{R_X} = 0 \quad (3)$$

$$F_{SP}\sin(\delta) - F_P\sin(\beta) + F_{G_5} + F_A - F_{R_Z} = 0 \quad (4)$$

A fourth equation is deduced from a sagittal X-ray image (Figure 3)

$$r_1 \times F_1 + r_2 \times F_2 + r_3 \times F_{SP} = 0 \quad (5)$$

Equations to determine the magnitudes of the two muscle groups, partial BW F_{G5} and the motion acceleration force F_A are

$$F_{SP} = -\frac{0.23L\cos\varphi}{r}(F_{G_5} + F_A) \quad (6)$$

$$F_P = \frac{\left(\frac{0.23sL\cos\varphi\sin(\Delta-\delta)}{r} - d_5\right)}{l\sin(\alpha - \beta)}(F_{G_5} + F_A) \quad (7)$$

Once the muscular forces are determined, the magnitude of F_R is given by

$$F_R = \sqrt{F_{R_X}^2 + F_{R_Z}^2} \quad (8)$$

Model according to Iglič. Iglič et al.[24] developed a static 3D model of the hip in order to evaluate the magnitude of the hip joint reaction force before and after the Chiari osteotomy (Figure 4). Like other hip models, the purpose of Iglič's model is to estimate the effects of orthopaedic interventions and to evaluate the biomechanical status of the hip with different femoral geometries.[24] F_R is determined by solving the equilibrium equations for forces and torques in the single-legged stance based on the individual measurements of the femoral and pelvic geometry in standard a.p. X-rays. The pelvis segment bears the partial BW (W_B–W_L) (= F_{G5}), where W_B is the total BW and W_L is the weight of the loaded leg. The weight of the leg is approximated to be 0.16 W_B.

Thus, the force and the moment equilibrium equations are

$$\sum_i F_i - F_R + (W_B - W_L) = 0 \quad (9)$$

$$\sum_i (r_i \times F_i) + a \times (W_B - W_L) = 0 \quad (10)$$

Where a = (0,0,a) is the moment arm of the force (W_B–W_L), r_i is the vector of the ith muscle force (F_i) application point. The orientation of each vector is from the origin of the coordinate system (= hip joint rotation centre) to the muscle origin point on the pelvis.[39] The model includes nine muscles that are classified into three groups according to their positions: anterior (α), middle (β), and posterior (γ). Furthermore, Iglič assumed that the force of each individual muscle (F_i) included in the model can be determined by the following vector

$$F_i = \sigma_i A_i e_i \quad (11)$$

$i = 1, \ldots, 9$

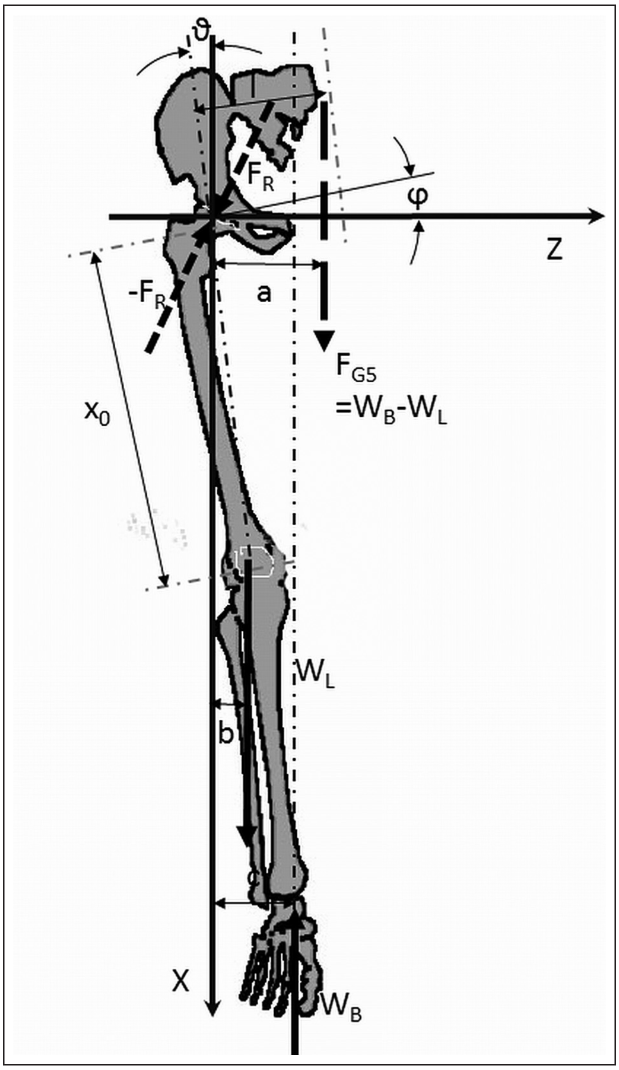


Figure 4. Iglič model (according to Iglič et al.[24]).

$$e_i = \frac{r_i' - r_i}{|r_i' - r_i|} \quad (12)$$

where A_i is the relative cross-sectional area of the ith muscle determined from the data of Johnston et al.,[15] σ_i is the average tension in the ith muscle and $e_i = (e_{ix}, e_{iy}, e_{iz})$ is the unit vector in the direction of the force of the ith muscle.[15,39] Each muscle is considered to act along the straight line connection between the origin point $r_i = (x_i, y_i, z_i)$ and the insertion point $r'_i = (x`_i, y`_i, z`_i)$. The rotation of the pelvis in the frontal plane around the *y*-axis is described by the angle (φ), while the rotation of the femur around the *y*-axis is described by the angle ϑ. For Iglič's model, the rotational angles are set as $\varphi = 0$ and $\vartheta = \arcsin(b/x_o)$, where x_o is the length of the femur and b is the *z*-coordinate of the moment arm of the force W_L. The length of the moment arm of the force (W_B–W_L) is determined from the *y*-component of the moment equilibrium equations

$$-W_B c + W_L b - M_Y = 0 \quad (13)$$

$$(W - B\,W_L)a + M_Y = 0 \quad (14)$$

$$a = \frac{W_B c - W_L b}{W_B - W_L} \tag{15}$$

where a is the z-coordinate of the moment arm a = (0,0,a), c is the z-coordinate of the moment arm of the ground reaction force $-W_B$, and M_Y is z-component of intersegmental moment

$$M = \sum_i (r_i \times F_i) \tag{16}$$

The moment arms b and c are expressed by the inter-hip distance l (b = 0.241 and c = 0.51).[39] To solve the equilibrium equations (9) and (10), and to determine the magnitude of the resultant force F_R, the average tensions (σ_i) in the particular muscle group are assumed to be equal: $\sigma_1 = \sigma_2 = \sigma_3 = \sigma_4 = \sigma_\alpha$, $\sigma_5 = \sigma_6 = \sigma_\beta$, $\sigma_7 = \sigma_8 = \sigma_9 = \sigma_\gamma$.[39] Hence, the unknown variables F_{Rx}, F_{Ry}, F_{Rz}, σ_α, σ_β, and σ_γ can be calculated.

Different from other mathematical hip models presented above, Iglič's model considers nine individual sets of muscle origins and insertion points (taken from Dostal and Andrews[40]). To calculate the hip resultant force for different individual patients, the individual muscle origin and insertion points, as well as the inter-hip distance, should be used if these data were available; otherwise the information given by Dostal and Andrews[40] should be used.

Material and methods

Comparison of different modelling approaches to estimate the resultant hip joint force F_R and its orientation

We performed a study aimed at examining how far the choice of the mathematical model influences the computational results of F_R, and to furthermore compare these results with *in-vivo* measurements published by Bergmann et al.[9,10,41] This comparison opens up the possibility to evaluate whether the calculated results are within an acceptable range of those found in *in-vivo* studies. Additionally, a multi-body simulation (MBS) investigation was carried out to compare the mathematical models with the simulation results from the 'AnyBody Modeling System™' software.

The investigation included the following clinically established models:[39,42,43] Pauwels',[17] Blumentritt,[22] Debrunner,[23] and Iglič et al.[24] They were all designed to calculate F_R in the single-legged stance phase based on patient-specific geometrical parameters derived from image data, together with patients' BW.

These models require standardized a.p. X-rays of the patient in a two-legged standing position for landmark detection (Figure 5). The differences in pelvic bend in single-legged stance phase versus two-legged upright position are neglected (only Iglič introduced a correction angle (φ), however, it is set to 0°[24]). The sensitivity of the different models related to this simplification requires careful verification.

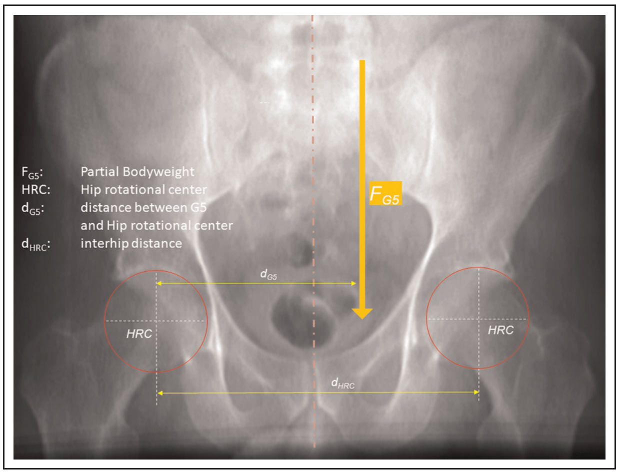


Figure 5. A digitally reconstructed radiograph (DRR) from a computed tomography (CT) dataset.

For our investigations, three digitally reconstructed radiographs (DRRs) in a.p.-direction were generated (Figure 5) based on computed tomography (CT) datasets from three patients (x21_x21, x12_x12, x8_x8) with a BW of 95 kg, 78 kg, and 90 kg. The DRR approximated the orientation of the pelvis for a patient standing on both legs according to the description of a radiographic pelvic overview in order to correct for differences in pelvic tilt (lying CT versus standing X-ray a.p. projection).[44,45]

All relevant parameters needed to calculate the amplitude and orientation of F_R are detected according to the specific modelling approach, as described for each model in 'Simplified biomechanical hip joint models'. In a quantitative comparative study with identical geometrical and anthropometrical parameters, F_R was computed by using these four models. The amplitude of F_R was expressed in per cent of BW and the orientation in terms of the angle θ as defined in Figure 1 for these three individual patient datasets. These model-based estimations have been compared with *in-vivo* telemetric measurement data.

The biomechanical behaviour of the hip can also be studied using direct *in-vivo* measurements.[7–10,38,46–53] Direct *in-vivo* measurement records the hip joint forces using instrumented prostheses, e.g. telemetrized total hip prosthesis by Bergmann et al.[10] This approach provides a basis to validate the mathematical models by comparing the calculated joint contact forces with *in-vivo* measurements.[54] The results from Bergmann were used as a primary reference in our investigations. The data are available online in the orthoLoad-database (www.orthoload.com).[9] An implanted instrumented prosthesis has an accuracy of 1%–2%.[41] Data on hip forces consist of tri-axial joint contact forces.[41]

The datasets for the single-leg stance and free walking without any constraints were the focus of our investigations.[55–57] Currently, three patient datasets (EBL, HSR, KWR; these are the anonymized patient datasets) for single-leg stance and one dataset for free walking without any external constraints (patient KWR) are available. However, the measurement method with an instrumented prosthesis can only be performed on a limited number of subjects and truly normal subjects cannot be studied. Currently, there is only a small number of datasets available. In each subject, only one type of prosthesis could be implanted and the effects of different interventions cannot be adequately explored.[38] Furthermore, even if there was one implanted device with various neck–shaft relationships in multiple patients so as to simulate interventions, the differences in forces observed would likely be overwhelmed by intersubject variables, such as gait patterns, muscle-activation patterns, and activity levels.[38]

To obtain the maximum hip load, the single-legged stance (static conditions) and the stance phase in planar gait (dynamic conditions) was considered, respectively. A reference value for the maximum hip load under static conditions was calculated by averaging the hip load values in a single-leg stance. Additionally, a reference value for the maximum hip load under dynamic conditions was calculated by averaging the first peak hip load values of three cycles (heel strike to toe off) determined in one single patient (among the three patients involved in the measurements under static conditions) in the database. Since the BWs of the three patients (EBL, HSR, KWR) from the orthoLoad-database are known, F_R was related to their respective bodyweights.

In the case of the orientation of F_R, it is also possible to estimate this parameter for individual patients based on orthoLoad-data. There was also a reference value for F_R under static conditions calculated by averaging the hip load values in single-leg stance (averaging 40 values around the highest peak) determined from three different patients from the data base. Additionally, a reference value for the maximum hip load under dynamic conditions was calculated by averaging the values of the highest gait cycle peaks and hip load values over three cycles (heel strike to toe off) determined in one single patient (KWR).

Additionally, we initialized a MBS-analysis using the AnyBody Modeling System. Computer based MBS-systems facilitate the modelling and analysis of 3D musculoskeletal systems.[3,5,11,58–61] The AnyBody Modeling System performs inverse dynamic modelling to simulate muscle and joint forces undergoing complex movements, taking mass inertia and contact forces into account.[16] In the context of the work reported here, this platform was used for a comparison between hip joint resultant forces obtained by the AnyBody simulation, and those obtained with various models described in the above mentioned literature.

In the MBS approach, the relevant input data, the geometrical values, and the anthropometrical parameters were extracted from the AnyBody *Standing Human Model* (Figure 6(b)) (Model repository V. 6.2), which approximates the whole human body. The relevant values were adopted into the different hip models of Pauwels,[17] Blumentritt,[22] and Iglič et al.[24] in order to compare all of them under the same biomechanical state and script environment. Debrunner's model was not considered because it was approximately the same as Pauwels' model. The hip joint resultant force value was determined according to each hip model's involved muscle forces, input body loads, and parameter values along with the AnyBody inverse dynamic operation. Next, the AnyBody *Standing Human Model*, which was assumed to be an exact and intricate human biomechanical model, was modified into a single-legged stance.

Sensitivity analysis of the Babisch-Layher-Blumentritt (BLB)-score

As the X-ray-based identification of the relevant anatomical landmarks could be one major source of error, we performed a sensitivity analysis of the so-called BLB-score based on Blumentritt's biomechanical hip model (see also Dell'Anna et al.[62]). This model is implemented in the mediCAD software (Hectec GmbH, Landshut, Germany) to plan the cup position.[22,42,63]

Nine anatomical landmarks (Figure 3) must be determined from an a.p. X-ray image of the hip so that the six patient-specific biomechanical parameters can be calculated.[43]

- P1 and P2 are only needed to define the position of the mid-sagittal plane.
- P3 is the pre-operative COR (hip centre of rotation).
- P4 approx. 3 cm distal from the lesser trochanter minor (direction of pull of the rectus femoris muscle towards the medial upper edge of the patella).
- P5 lateral, most proximal edge of the trochanter major.
- P6 most cranial edge of the sclerotic area.
- P7 spina iliaca anterior inferior.
- P8/P9 most lateral/cranial point of the wing of the ilium.

However, the varying quality of X-ray images and the inter- and intra-variability of the medical expert users may lead to inaccuracies in landmark identification (see also Fieten et al.[64]). The extent and effect of these inaccuracies on the biomechanical score are of interest with respect to the robustness of the score and the reliability of the THA planning. In this context it might be of special interest, whether there are landmarks with a significantly higher influence than other landmarks and whether these landmarks can be identified accurately enough in clinical routine. To answer these questions we prepared a sensitivity analysis of the BLB-score regarding inaccuracies in landmark identification.[65]

Two DRRs were built from the CT-datasets of two patients, showing the hip in an a.p. position. DRRs were used in order to make sure that the required X-ray parameters, such as the focus of central X-ray beam, were complied with and the X-ray projections could be directly related to CT-based 3D reconstructions. These two DRRs were presented to 11 orthopaedic surgeons, who determined the anatomical landmarks several times in each image. Altogether each image was presented 45 times, leading to point clouds of 45 points for each landmark.

For statistical reasons the six biomechanical parameters, instead of the biomechanical score, were chosen as output variables of the analysis. A centre point was calculated for all of the point clouds of each landmark. Then one landmark was shifted stepwise (the others were fixed) and its impact on the BLB-score was evaluated.

Results

Amplitude and orientation of F_R, MBS-approach

The results are summarized in Tables 1–5. Table 1 shows the results for the amplitude of F_R. The mean value for the single-leg stance is 2.69 of BW (269% BW). The investigations for single-leg stance were made post operation (EBL 1.1 month, HSR 10 month, KWR 8 month). There is only one patient dataset for free walking with the value of 2.12 BW (212% BW) available in the orthoLoad-database. In the case of dynamic investigation (5 month post operation) the patient KWR had a slightly increased BW.

Table 2 shows the computational results for the amplitude of F_R. The results of Pauwels 2.74 BW (274% BW), Debrunner 2.54 BW (254% BW), and Iglič 2.42 BW (242% BW) are approximately in the same range. Only Blumentritt's result of 6.45 BW (645% BW) shows an increased outcome in comparison the other models. Note that the X-ray data we used for F_R-computation are not the data from the patients of the orthoLoad-database.

Table 3 shows the results for the orientation of F_R calculated from the datasets of the orthoLoad-database. The mean value for a single-leg stance is 18.47°, and in the case of free walking it is 15.71°.

The computed results for each model are listed in Table 4. The results of Pauwels (18.64°), Debrunner (13.09°), Iglič (9.15°), and Blumentritt (5.95°) differ from each other. The computational results showed large variations for F_R orientation, which tends to depend more on the model used than on patient-specific parameters.

Table 5 shows the simulation result for the AnyBody *Standing Human Model* 3.20 BW (320% BW) in comparison to the computed results of Pauwels 3.28 BW (328% BW), Iglič 2.45 BW (245% BW), and Blumentritt 5.34 BW (534% BW). Blumentritt's result also shows an increased outcome in comparison with the other models, as mentioned in the case of computational results in Table 2. In Figure 6(a) the results are listed for simulated knee flexion. The resultant joint force decreased during increasing knee flexion.

Table 1. Results for the amplitude of F_R for the orthoLoad-data.

orthoLoad			
Patient	Maximum value F_R [N]	BW [N]	F_R/BW
EBL	1967.19	650.00	3.03
HSR	2000.43	870.00	2.30
KWR	1889.75	690.00	2.74
Mean value	1952.46	736.67	2.69
KWR (free walking)	1565.33	740.00	2.12

BW: body weight; EBL, HSR, KWR are the patient datasets taken from the orthoLoad-database

Sensitivity analysis

The landmark detection showed strong scattering in most of the landmarks (Figure 7), except in landmarks P2, P3, and P6, intra-individually as well as inter-individually. Regarding the influence of inaccuracies in landmark identification for the computation of the BLB-score, the sensitivity analysis showed large differences between the anatomical landmarks. Landmark P7 showed the highest influence by far because correct identification seems to be problematic (Figure 7). However, landmarks P3, P5, and P4 showed a distinct influence as well. In contrast, the other landmarks only seem to have a moderate impact on the BLB-score computation.

Discussion

Amplitude and orientation of F_R, MBS-approach

Amplitude and orientation of F_R. By comparing the *in-vivo* amplitude values of F_R obtained under dynamic conditions, i.e. gait, with the *in-vivo* values of the same

Table 2. Results for the amplitude of F_R of the mathematical models.

Patient data								
Model	Pauwels		Debrunner		Blumentritt		Iglič	
	F_R	F_R/BW	F_R	F_R/BW	F_R	F_R/BW	F_R	F_R/BW
x21_x21	2048.27	2.16	2151.20	2.26	6130.07	6.45	2312.76	2.43
x12_x12	2227.22	2.86	2150.80	2.76	5182.74	6.64	1969.34	2.52
x8_x8	2880.24	3.20	2347.45	2.61	5632.79	6.25	2078.53	2.31
Mean value	2385.24	2.74	2216.48	2.54	5648.53	6.45	2120.21	2.42

BW: body weight.

patient obtained under static conditions, it could be recognized that the dynamic values are even smaller than the static measurement results. Unlike the other mathematical hip models for the single-leg stance, Blumentritt's model has an additional motion acceleration force F_A and a knee flexion of 20°. These two factors might lead to the overestimation of F_R, at least for slow walking. This tendency is amplified by the addition of the rectus femoris muscle force to the abductor muscle force.

Table 3. Orientation of F_R of individual patient datasets from the orthoLoad-data.

orthoLoad			
Patient	F_R [°]		F_R [°]
EBL	25.20	–	–
HSR	17.09	–	–
KWR	13.12	free walking	15.71
Mean value	18.47		15.71

EBL, HSR, KWR are the patient datasets taken from the orthoLoad-database

Table 4. Orientation of F_R of individual patient data based on DRRs and calculated from orthoLoad-data.

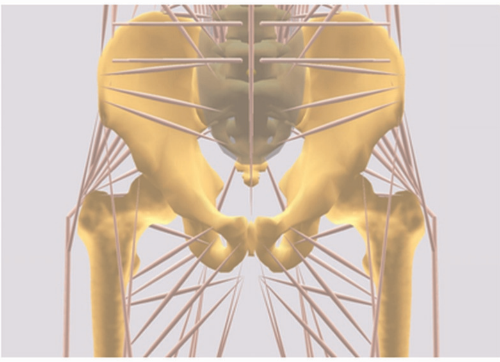
Patient data F_R [°]				
Model	Pauwels	Debrunner	Blumentritt	Iglič
x21_x21	18.26	12.37	5.18	9.24
x12_x12	17.31	12.59	6.14	9.20
x8_x8	20.34	14.30	6.52	9.01
Mean value	18.64	13.09	5.95	9.15

Table 5. Simulated joint resultant forces using AnyBody *Standing Human Model* (BW = 738.931 N) input data.

AnyBody Modeling System™				
Models	Pauwels	Blumentritt	Iglič	Standing model
Muscle forces [F_M/BW]	F_M = 2.52	F_p = 1.56 F_{sp} = 2.80	F_M = 1.67	–
F_R/BW	3.28	5.34	2.45	3.20
[%] Difference in F_R with the standing model	+ 2.50	+ 66.88	–23.44	0
[%] Difference in F_R with the orthoLoad average value	+ 21.93	+ 98.88	–8.92	+ 18.96

BW: body weight.

AnyBody Modeling System™



Condition	Joint Resultant Force [F_R/BW]
0° knee flexion with complete muscles (Ideal)	3.20
0° knee flexion with only abductor muscle group muscles	3.01
20° knee flexion with complete muscles	2.49
20° knee flexion with only abductor muscle group muscles	2.67

(a)

Body Mass	Body Height	Single leg Mass	Partial Body Mass	Femur Length	Inter Hip Distance
75.35 kg = 738.931 N	1.8 m	12.075 kg = 118.415 N	63.275 kg = 620.516 N	0.4558 m	0.16 m

(b)

Figure 6. (a) Different model conditions and their simulation results; (b) parameter of the AnyBody Modeling System.

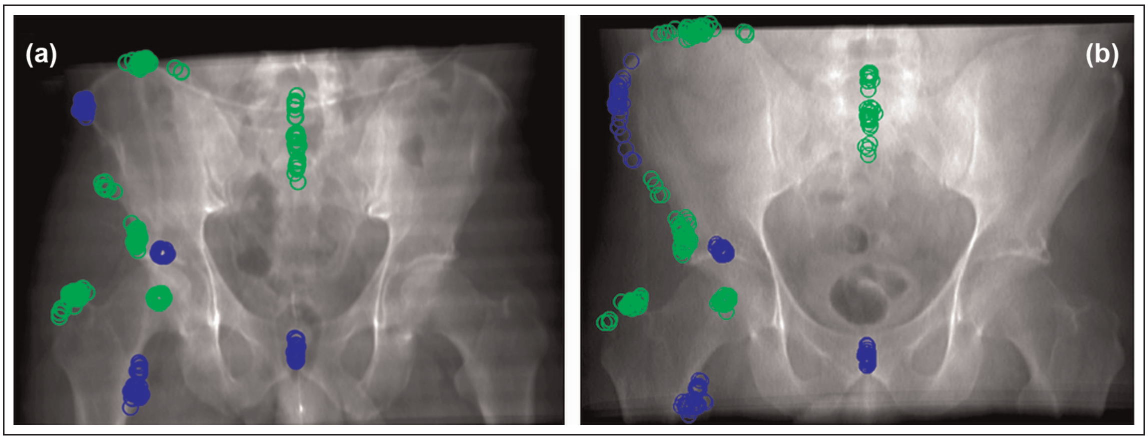


Figure 7. DRRs point clouds (a) and (b).

One can conclude that the assumptions made for the Blumentritt model must be carefully reviewed, as deviations from *in-vivo* measurement data become apparent. It should also be noted that the Blumentritt model was developed so as to evaluate the hip joint biomechanically under the aspect of pelvic development, resulting in a biomechanical score for the hip joint (BLB-score).[22,25]

The influence of the pelvic bend[66] during gait was only considered by Iglič's model by the angle φ (Figure 4). The other models ignore this parameter. However, Iglič also sets the angle φ to zero.[24]

Regarding the values for orientation of F_R, the mean values of Debrunner and Iglič are slightly smaller, whereas Pauwels is slightly higher than the mean value of the static in-vivo results. Only Pauwels result is higher than the in-vivo dynamic result. A possible explanation for these results could be inaccuracies in the data acquisition from the DRR. The computational results showed large variations for F_R orientation, which tend to depend more on the model used than on patient-specific parameters. By comparing the *in-vivo* values obtained under dynamic conditions, i.e. gait, with the static *in-vivo* values of the same patient, it could be recognized that the static values are slightly smaller than the dynamic result. But both are nearly in the same range as the mathematical models. In contrast, Blumentritt's outcomes have the largest deviation from the other models as well as from the *in-vivo* data (static and dynamic conditions). Blumentritt used the weight bearing surface as a reference, perpendicular to the longitudinal axis. He postulated that a valid and optimal orientation of F_R is perpendicular on it, respectively parallel to the z-axis. This approach for validation is questionable because a survey on the results shows that in the three included and analysed DDR's the orientations are not parallel to the longitudinal axis.

Among the limitations of our study is the fact that the orthoLoad-database offers only a small number of patient datasets. Only one data set is available that makes a comparison between static (single-leg stance) and dynamic (free planar gait) *in-vivo* measurement data possible for the same patient. Furthermore, the individual anatomic geometry data of the patients included in the database are not revealed.

MBS-approach. The magnitude of F_R for Pauwels' model was the only value within a 5% difference in comparison with the result of the AnyBody *Standing Human Model.* The simulated joint resultant force of Blumentritt's model was more than five times the total BW.

When comparing with the average direct-measured joint resultant force in a single-legged stance from the orthoLoad research group (average static 2.69 BW), Iglič's model, with a percentage difference of 8.92%, shows the only satisfactory result. Once more, the simulated joint resultant force from Blumentritt's model was not realistic, having a 98.88% difference. Furthermore, the magnitude of F_R from the AnyBody *Standing Human Model* in different posture conditions was listed in Figure 6(a). It was ideal to exam the hip joint resultant force F_R including all muscles in the human body for the simulation.

As most of the mathematical hip models[17,23] only considered the abductor muscle group and sometimes the rectus femoris muscle,[22] it was noteworthy to observe how the AnyBody *Standing Human Model* would behave in similar conditions to the mathematical models. As a result, the difference between the ideal to the condition that included only the abductor muscle group was less than 6%. It was also interesting to see that, when the knee flexion increased by 20°, the joint resultant force magnitude was lowered.

Sensitivity analysis

The sensitivity method only allows for comparative statements and is only based on two patient datasets. However, differences in the impact of inaccuracies in landmark identification on the score could be detected and should be further investigated. The extent of scattering observed in the clinical trial verified the necessity to perform a sensitivity analysis of the score. Moreover, the strong scattering could not only originate from a slightly worse image quality of the generated DRRs compared with real X-ray images. Obviously, an unambiguous identification of most landmarks in an X-ray (2D) image is hardly possible and the investigation (see also Fieten et al.[64,67]) of alternative identification methods, such as CT (3D-model of the pelvis) or ultrasound, are strongly advised for P7, P5 and P4 landmarks, which showed a high impact on the score. Moreover, landmark P4 could be identified more accurately by using a second X-ray image a.p. that includes the hip and patella. Landmark P3 is identified relatively precisely. The variability of the other landmarks seems to be noncritical since they only show a weak influence on the score. In the future, the influence of interactive landmark selection and also the influence of the X-ray parameters (standardization of a.p. X-ray imaging) on the biomechanical score should be further investigated.

Summary

Knowledge of biomechanical conditions of an individual hip is important for several reasons: understanding the function of the normal and diseased hip joints, diagnosing the condition of the hip, planning better treatments, evaluating the effects of treatments, optimizing implant design and performance, and improving the treatment outcomes.

Regarding the magnitude of F_R, we showed that the Pauwels', Debrunner's and Iglič's models are all within the same range compared with the *in-vivo* data provided by Bergmann et al.[9] The large deviation of the results obtained from using Blumentritt's model might be partly explained by the additional load F_A within his model. This load F_A is added to the input parameter BW in order to consider the dynamic effects of the human gait. Such a dynamic supplemental load was not applied in the other static model investigations of the single-leg stance. The results obtained with *in-vivo* measurement data in the static case are best approximated by using Pauwels' model. In the *in-vivo* measurements of Bergmann et al.,[9,10,41,68] the hip resultant only reached the level up to $6 \times$ BW (as calculated by the Blumentritt model) in case of fast running or up to $8 \times$ BW during stumbling – which however is not considered in the modelling approach for hip surgery planning.

In the case of an MBS-approach we showed that there was a discrepancy between the AnyBody *Standing Human Model* and the instrumented measurement results obtained by Bergmann. This could have several reasons. For example, one reason could be that the patients of the *in-vivo* measurements had undergone an arthroplasty surgery, which would have an influence on the gait and on the measurement results, respectively. Hence, it might cause a difference between the ranges of the forces compared with normal subjects. The adaption of the AnyBody *Standing Human Model* to the anatomical geometry of the patients of the orthoLoad-database was not possible, because the information about their individual geometrical and functional anatomy was not available.

Our sensitivity analysis of the BLB-score demonstrated problems with erroneous landmark detection. The sensitivity analysis proposed only allows comparative statements and was based on only two patient datasets. However, differences in the impact of inaccuracies of landmark identification on the BLB-score could be detected.

Conclusion and outlook

It is recommendable that further research be performed on implementing biomechanical hip models during the planning phase of computer-assisted THA. Sensitivity analyses and parameter studies for different mathematical models using a MBS system demonstrate that there is a strong need for further investigations.

The use of dynamic hip models is often too demanding and complicated during the daily clinical practice. Therefore, static mathematical models, as proposed by Pauwels, are preferred owing to their simplicity.[24] Indeed, Pauwels' model has demonstrated the best efficiency and accuracy in estimating the magnitude of the hip joint resultant force in two dimensions. For a 3D estimation of the hip joint resultant force, Iglič's model has the minimum discrepancy from the directly measured data.

With the knowledge of amplitude and orientation of F_R, it would be possible, to optimize the load transmission on an artificial hip joint and increase the lifetime of the implant.[69] In regards to mathematical models, future steps would include the transfer of patient-specific geometrical anatomical data from the orthoLoad-database to the different models and the direct comparison of the computed results with the *in-vivo* measurements for each individual patient. Additionally, investigation of pelvic orientation and the influence on the results would be considered in the future.

Our aim is to develop a simulation and planning tool enabling the surgeon to explore and optimize the biomechanical consequences of a surgery in a computer environment before stepping into the operating room. A highly individualized planning and surgical procedure that takes into account patient-specific constraints would help to meet the requirements presented by increasingly younger and/or heavier patients.[61]

Funding

This work has been funded in part by the German Ministry for Education and Research (BMBF) within the framework of the orthoMIT project under grant No. BMBF 01EQ0402/BMBF 01IBE02C.

Conflict of interests

The Authors declare no conflict of interest.

References

1. Yoshida H, Faust A, Wilckens J, et al. Three-dimensional dynamic hip contact area and pressure distribution during activities of daily living. *J Biomech* 2006; 39(11): 1996–2004.
2. Crowninshield RD, Johnston RC, Andrews JG, et al. A biomechanical investigation of the human hip. *J Biomech* 1978; 11(1–2): 75–77.
3. Hashimoto N, Ando M, Yayama T, et al. Dynamic analysis of the resultant force acting on the hip joint during level walking. *Artif Organs* 2005; 29(5): 387–392.
4. Statistisches Bundesamt – Fallpauschalenbezogene Krankenhausstatistik (DRG-Statistik), 2011.
5. Heller MO, Bergmann G, Kassi J-P, et al. Determination of muscle loading at the hip joint for use in pre-clinical testing. *J Biomech* 2005; 38(5): 1155–1163.
6. O'Toole RV, Jaramaz B, DiGioia AM, et al. Biomechanics for preoperative planning and surgical simulations in orthopaedics. *Computers Bio Med* 1995; 25(2): 183–191.
7. Davy DT, Kotzar GM, Brown RH, et al. Telemetric force measurements across the hip after total arthroplasty. *J Bone Joint Surg Am* 1988; 70(1): 45–50.
8. Brand RA, Pedersen DR, Davy DT, et al. Comparison of hip force calculations and measurements in the same patient. *J Arthroplasty* 1994; 9(1): 45–51.
9. Bergmann G, Graichen F, Rohlmann A. Hip joint loading during walking and running, measured in two patients. *J Biomech* 1993; 26(8): 969–990.
10. Bergmann G, Deuretzbacher G, Heller M, et al. Hip contact forces and gait patterns from routine activities. *J Biomech* 2001; 34(7): 859–871.
11. Heller MO, Schroeder JH, Matziolis G, et al. Musculoskeletal load analysis. A biomechanical explanation for clinical results–and more? *Orthopaede* 2007; 36(3): 190–188.
12. Rohrle H, Scholten R, Sigolotto C, et al. Joint forces in the human pelvis-leg skeleton during walking. *J Biomech* 1984; 17(6): 409–424.
13. Pedersen DR, Brand RA and Davy DT. Pelvic muscle and acetabular contact forces during gait. *J Biomech* 1997; 30(9): 959–965.
14. Seireg A and Arvikar RJ. The prediction of muscular load sharing and joint forces in the lower extremities during walking. *J Biomech* 1975; 8(2): 89–102.
15. Johnston RC, Brand RA and Crowninshield RD. Reconstruction of the hip. a mathematical approach to determine optimum geometric relationships. *J Bone Joint Surg Am* 1979; 61(5): 639–652.
16. www.anybodytech.com. AnyBody Tutorial 5.0, V. 5.0 Edition, 2010.
17. Pauwels F. *Atlas zur Biomechanik der gesunden und kranken Huefte – Prinzipien, Technik und Resultate einer kausalen Therapie*. Berlin/Heidelberg/New York: Springer Verlag, 1973.
18. Pauwels F. Die Bedeutung der Muskelkraefte fuer die Regelung der Beanspruchung des Roehrenknochens waehrend der Bewegung der Glieder. *Anatomy and Embryology* 1951; 115: 327–351.
19. Pauwels F. Die Bedeutung der Bauprinzipien der unteren Extremitaet fuer die Beanspruchung des Beinskeletes. *Anatomy and Embryology* 1950; 114: 525–538.
20. Pauwels F. Die Bedeutung der Bauprinzipien des Stuetz- und Bewegungsapparates fuer die Beanspruchung der Roehrenknochen. *Anatomy and Embryology* 1948; 114: 129–166.
21. Brinckmann P, Frobin W and Hierholzer E. Stress on the articular surface of the hip joint in healthy adults and persons with idiopathic osteoarthrosis of the hip joint. *J Biomech* 1981; 14(3): 149–156.
22. Blumentritt S. The relationship between the gait of humans and the hip joint structure in the frontal plane. *Gegenbaurs Morphol Jahrb* 1990; 136(6): 677–693.
23. Debrunner HU. Biomechanics of the hip joint. I. A new model for the calculation of the forces in the hip joint. *Z Orthop Ihre Grenzgeb* 1975; 113(3): 377–388.
24. Iglič A, Srakar F, Antolič V, et al. Mathematical analysis of chiari osteotomy. *Acta Orthop Jugosl* 1990; 20(2–3): 35–39.
25. Blumentritt S. Biomechanische Bauprinzipien des menschlichen Hueftgelenkes in der Frontalebene. *Gegenbaurs morphol Jahrb L*eipzig 1988; 134: 221–240.
26. Hamacher P and Roesler H. Results of calculations on the size and direction of resultant hip-joint forces for the individual case. *Arch Orthop Unfallchir* 1972; 72(1): 94–106.
27. Hamacher P and Roesler H. Die Berechnung von Groeße und Richtung der Hueftgelenksresultierenden im Einzelfall. *Archives of Orthopae Trauma Surg* 1971; 70(1): 26–35.
28. Kummer B. Biomechanical principles of the statics of the hip joint. critical appraisal of a new theory. *Z Orthop Ihre Grenzgeb* 1986; 124(2): 179–187.
29. Kummer B. Die Beanspruchung des menschlichen Hueftgelenks. *Anatomy and Embryology* 1968; 127(4): 277–28.
30. Kummer B. The clinical relevance of biomechanical analysis of the hip area. *Z Orthop Ihre Grenzgeb* 1991; 129(4): 285–294.
31. Moeser M and Hein W. Kraefte an der Huefte – Das Untergurtmodell Teil 1: Kritik am Pauwels-Modell – der Zweibeinstand. *Beitr Orthop Traumatol* 1987; 34(2): 83–92.
32. Moeser M and Hein W. Kraefte an der Huefte – Das Untergurtmodell Teil 2: Der Einbeinstand. das Turmkranprinzip. *Beitr Orthop Traumatol* 1987; 34(4): 179–189.
33. Tschauner C, Hofmann S and Czerny C. Hip dysplasia. Morphology, biomechanics and therapeutic principles with reference to the acetabular labrum. *Orthopaede* 1997; 26(1): 89–108.
34. Paul JP. Biomechanics. The biomechanics of the hip-joint and its clinical relevance. *Proc R Soc Med* 1966; 59(10): 943–948.
35. Paul JP. Force actions transmitted by joints in the human body. *Proc R Soc Lond B Biol Sci* 1976; 192(1107): 163–172.
36. Maquet P. Biomechanics of hip dysplasia. *Acta Orthop Belg* 1999; 65(3): 302–314.
37. Genda E, Iwasaki N, Li G, et al. Normal hip joint contact pressure distribution in single-leg standing - effect of gender and anatomic parameters. *J Biomech* 2001; 34(7): 895–905.
38. Brand RA. Hip osteotomies: A biomechanical consideration. *J Am Acad Orthop Surg* 1997; 5(5): 282–291.
39. Iglič A, Kralj-Iglič V, Daniel M, et al. Computer determination of contact stress distribution and size of weight bearing area in the human hip joint. *Comput Methods Biomech Biomed Engng* 2002; 5(2): 185–192.

40. Dostal WF and Andrews JG. A three-dimensional biomechanical model of hip musculature. *J Biomech* 1981; 14(11): 803–807.
41. www.orthoload.com; www.jwi.charite.de, October 2011.
42. Babisch J, Layher F, Venbrocks R, et al. Biomechanisch fundierte Hueftoperationsplanung mit Hilfe des Softwaremoduls Endomap. *Eelctromedia* 2002; 70(1): 39–46.
43. HECTEC. *mediCAD user's manual* V2.06, HECTEC GmbH, Landshut, Germany, v2.06 edition.
44. Babisch J, Layher F and Amiot LP. The rationale tilt of tilt-adjusted acetabular cup navigation. *J Bone Joint Surg Am* 2008; 90: 357–365.
45. Wirth CJ, Zichner L and Tschauner C. *Orthopaedie und Orthopaedische Chirurgie – Becken, Huefte*. Georg Thieme Verlag Stuttgart, 2004.
46. English TA and Kilvington M. In vivo records of hip loads using a femoral implant with telemetric output (a preliminary report). *J Biomed Eng* 1979; 1(2): 111–115.
47. Hodge WA, Carlson KL, Fijan RS, et al. Contact pressures from an instrumented hip endoprosthesis. *J Bone Joint Surg Am* 1989; 71(9): 1378–1386.
48. Kotzar GM, Davy DT, Goldberg VM, et al. Telemeterized in vivo hip joint force data: a report on two patients after total hip surgery. *J Orthop Res* 1991; 9(5): 621–633.
49. Taylor SJ, Perry JS, Meswania JM, et al. Telemetry of forces from proximal femoral replacements and relevance to fixation. *J Biomech* 1997; 30(3): 225–234.
50. Taylor SJ, Walker PS, Perry JS, et al. The forces in the distal femur and the knee during walking and other activities measured by telemetry. *J Arthroplasty* 1998; 13(4): 428–437.
51. Bombelli R. *Osteoarthritis of the hip – Classification and pathogenesis – The role of osteotomy as a consequent therapy*. 2nd ed. Berlin/Heidelberg/New York: Springer Verlag, 1983.
52. Stansfield BW and Nicol A.C. Hip joint contact forces in normal subjects and subjects with total hip prostheses: walking and stair and ramp negotiation. *Clin Biomech* 2002; 17(2): 130–139.
53. Stansfield BW, Nicol AC, Paul JP, et al. Direct comparison of calculated hip joint contact forces with those measured using instrumented implants. an evaluation of a three-dimensional mathematical model of the lower limb. *J Biomech* 2003; 36(7): 929–936.
54. Henninger HB, Reese SP, Anderson AE, et al. Validation of computational models in biomechanics. *Proc IMechE, Part H: J Engineering in Medicine* 2010; 224(7): 801–812.
55. Eschweiler J, Fieten L, Belei P, et al. A comparison of different biomechanical hip-models. In: *Proceedings of the 10th annual meeting of CAOS international*, 2010, pp.518–521.
56. Eschweiler J, Fieten L, Belei P, et al. Vergleichende Analyse bekannter biomechanischer Hueftmodelle. *Biomedizinische Technik* 2010; 55(Sup1): Jahrestagung Biomedizinische Technik, Walter de Gruyter.
57. Eschweiler J, Fieten L, Schmidt F, et al. Comparison of the orientation of the resultant force of different biomechanical hip models. In: *Proceedings of the 11th annual meeting of CAOS international*, London, 2011.
58. Delp SL and Maloney W. Effects of hip center location on the moment-generating capacity of the muscles. *J Biomech* 1993; 26(4–5): 485–499.
59. Lengsfeld M, Bassaly A, Boudriot U, et al. Size and direction of hip joint forces associated with various positions of the acetabulum. *J Arthroplasty* 2000; 15(3): 314–320.
60. Amankwah K, Triolo R, Kirsch R, Audu M. A model-based study of passive joint properties on muscle effort during static stance. *Journal of Biomechanics*. 2006; 39(12): 2253–2263.
61. Rasmussen J. Musculoskeletal analysis of navigated total hip arthroplasty. *Bonezone* 2009; 8(3): 86–89.
62. Dell'Anna J, Fieten L, Eschweiler J, et al. Sensitivity analysis of a biomechanical score for cup positioning in total hip arthroplasty. In: *Proceedings of the 10th annual meeting of CAOS international*, 2010, pp.364–366.
63. Babisch J, Layher F and Venbrocks R. Computer assisted planning and navigation using the Navitrack and MediCAD. In: *SystemNavigation and robotics in total joint and spine surgery*. Springer Edition, 2004.
64. Fieten L, Eschweiler J, Kabir K, et al. Reliability of mental 2d/3d matching for 3d cup positioning based on preoperative 2d planning in defect hips. In: *Proceedings of the 11th annual meeting of CAOS international*, London, 2011.
65. Dell'Anna J, Fieten J, Eschweiler J, et al. Sensitivitaetsanalyse eines biomechanischen Scores zur Pfannenpositionierung bei einer totalen Hueft-Arthroplastie. *Biomedizinische Technik* 2010; 55(Sup1): Jahrestagung Biomedizinische Technik, Walter de Gruyter.
66. Saunders JB, Inman VT and Eberhardt HD. The major determinants in normal and pathological gait. *J Bone Joint Surg Am* 1953; 35-A(3): 543–558.
67. Fieten L, Eschweiler J, Kabir K, et al. 2D vs. 3D landmark localization for cup positioning in THR based on biomechanical hip models. In: *Proceedings of the 11th annual meeting of CAOS international*. London, 2011.
68. Bergmann G, Graichen F and Rohlmann A. Hip joint contact forces during stumbling. *Langenbecks Arch Surg* 2004; 389: 53–59.
69. Thomas W and Schug M. Significance of the position of the endoprosthesis acetabular cup from the biomechanical and clinical viewpoint–recommendations for a classification. *Biomed Tech* (*Berl*) 1994; 39(9): 222–226.