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Musculoskeletal load analyses

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In biomechanics, a great deal of research has been conducted in an effort to characterize the mechanics and interaction of the soft and hard tissue structures. Unfortunately, many investigators tend to treat the muscles and bones as separate and independent structures. However, in order to fully understand how loads are transmitted within the body the mechanical interaction of these structures must be considered. An example of the complex inter-relation of the soft and hard tissue structures are the long bones of the upper and lower extremities. For the lasting function of an artificial joint or for the reconstruction of fractured bones the mechanical loading of the biological tissue is essential: Implants may fail under peak or fatigue loads and the biological healing or adaptation process is triggered by mechanical stimuli.
Therefore, numerous studies have been performed to examine the mechanical conditions of the musculo-skeletal system of the upper and lower limb. Only seldom these investigations incorporated the bones with their muscular and ligamentous structures. Because of its importance in the musculo-skeletal system, a complete understanding of the load situation in a bone is of particular interest.
In the first part of this chapter, it is explained how to calculate the load conditions in long bones of the lower limb. The unknowns associated with these calculations are the muscle, ligament and joint contact forces; in addition the exact anatomy is seldom known. Joint contact and ligament forces may be determined from in vivo and in vitro experiments. Muscle forces can be estimated by mathematical models, direct measurements requires the monitoring of the electromyographic activity. The literature on these measurements and calculations will be reviewed; the techniques used in the computation of internal loads are formulated.

Key words: *musculo-skeletal loading, internal loads in bones, optimisation, in vivo loading.*

1. Experimental and analytical analysis

The experimental and analytical analysis of a loaded bone represents a common procedure in biomechanics. Investigations involving the fixation of an implant in the bone, implant design or fracture fixation require knowledge of the physiological loading of the bone and implant. In joint replacement operations, it may be questioned to what degree a surgical approach modifies the load distribution within a long bone and may lead to bone loss or reduced tissue function. In all these instances, knowledge of the forces and moments acting within a long bone would facilitate appropriate medical treatment options.

The load situation within a bone has been a subject of study in research for many years. As early as 1638, Galileo studied the mechanics of long bones and analysed the gross anatomical structure of the thigh [1]. The basis of his work and further descriptive analysis assumed a relationship between mechanical principles and the anatomical shape of the femur. Surprisingly, the inner structure, not the macroscopic anatomy, has been the main focus in biomechanical analyses of bone. Based on Meyers description of the trabecular architecture in the proximal femur [2], Wolff postulated that bone structure corresponds to the load path. The trabecular structures were found to be regularly orientated and aligned to form a sophisticated pattern. Wolff formulated that form follows function [3] and quoted the work of Culmann (1866) in which the advantages offered by the orientation of the trabecular structure could be interpreted by mechanical means. Assuming that bone was an optimum structure, it was obvious to interpret that the bony architecture resulted from the mechanical influences that occurred during evolution. In addition, Wolff found that fully healed misaligned fractures also obeyed mechanical rules. In his work from 1892, Wolff stated explicitly that bony structures were not solely pre-determined by genetic factors but also by adaptation to mechanical load situations ("Law of Bone Remodelling" or Wolff's Law [4]).

1.1. Historical review of musculoskeletal analysis

Although Wolff described the adaptation of bone to mechanical situations, it was Koch, who first tried to quantify the mechanical situation within a bone by calculating the stresses and strains [5]. In his publication, Koch thoroughly analysed the anatomy of the femur by calculating the cross-sectional area and moment of inertia at seventy-five locations. The cross-sections were aligned perpendicular to a line between the centroids of the knee and femoral head centre. Using the geometrical data of the bone and an assumed set of material properties, Koch was able to calculate the forces acting in line with the long axis of the bone (axial forces) and the ones acting perpendicular to the long axis (shear forces) as well as the bending moments and principle stress lines in the femur. Koch used for his calculation a method known as "beam theory". Koch assumed a hip joint contact force of 445 N ($\approx$ 0.7 BW) during walking and running. For this joint load, the axial force in the femur increased from 177 N at the hip to 445 N distal of the lesser trochanter. A maximum shear force of 409 N was present at the femoral head and decreased towards the lesser trochanter; the maximal bending moments occurred at the level of the lesser trochanter with values of up to 18 Nm ($\approx$ 0.03 BWm). Surprisingly, the direction of his computed principle stresses agreed well with the descriptions of trabecular architecture made by Wolff in 1892. Koch's analysis was only capable of representing femoral curvatures in the frontal plane and neglected to include muscle activities. The later led to an underestimation of the joint contact force and, consequently, the femoral loads. Nevertheless, his work is considered the classical approach to femoral stress calculation that opened the door for numerous research studies on this subject. Shortly after Koch, Grunewald [6] and Marique [7] published similar works on the mechanical environment within the femur. Evans and Lissner [8] and later Frankel [9] used a brittle stress coating to experimentally investigate the stress situation of the loaded bone. Evans and Lissner found, in good agreement to Koch's calculations, stress maxima in the femoral head and condylar regions. Stresses in the diaphyseal region were found to be reasonably constant; no muscle forces were considered.

Pauwels was one of the first researchers to include the effect of muscles on femoral loading in his analytical analyses [10] and experiments [11]. Due to the experimental set-up used in his photoelastic investigations, the three-dimensional characteristics of the long bones were ignored. Nevertheless, Pauwels stated that a bone is loaded by a bending moment superimposed by compression [12]. He was able to detect tension (lateral) and compression (medial) patterns in the femoral neck and shaft regions. Further, Pauwels showed that tension in a band between the greater trochanter and femoral condyles significantly reduced the bending moments.

Neglecting the findings of Pauwels, Torodis [13] assumed that the effect of the muscle loads on the mechanical behaviour of a bone was not as important as that of the body weight. Rybicki used the anatomical description of the femur described by Koch, to analyse the femoral stresses in the one-legged stance phase of gait [14]. Rybicki's calculations included the joint contact force (2318 N $\approx$ 3.6 BW), the hip abductors (m. gluteus medius and m. gluteus minimus, 1592 N) and the iliotibial band. The later was modelled as a tension band between the tibial epicondyles and iliac crest, gliding on, but not connected to, the greater trochanter. The muscle forces were taken from calculations based on the work of Inman [15] and the joint forces from studies conducted by Bresler and Frankel [16]. The effects associated with a wide range of forces in the iliotibial band (0 ... 1557 N) were investigated using both beam theory and two-dimensional finite element modelling. Similar to Pauwels, Rybicki's results indicate a reduction of the maximum stresses (70–77% of the original values) within the bone due to iliotibial activity. The author argued that including more muscles would most likely result in further reduction of the bending moments and, hence, the stresses acting in the bone. Rybicki concluded, that although finite element models were more accurate in the regions of the greater trochanter and femoral head (beam theory overestimated the stresses by a factor of two), the results were quite comparable to those of beam theory in the femoral shaft region (factor < 1.5). Although Rybicki considered the mm. abductores and the iliotibial band, no other muscular influences were analysed (mm. adductores, mm. vasti). His model was basically two-dimensional representing only one load situation, single leg stance. Furthermore, no distinctions were made among the six load components or between the resulting stress directions. Therefore, the conclusions drawn from this study are quite limited in general implications.

In 1976, Ghista et al. published a rather complete mathematical description of the internal stresses in a bone during gait [17]. The muscle forces were determined by application of an inverse dynamic calculation of the resultant loads at the instantaneous joint centres. To satisfy the equilibrium conditions for a limb segment, Ghista simplified the model such that only three muscles and three joint contact forces were active at a time. The method allowed computation of the internal stresses of the bone using the force data and geometric properties of the femur. Unfortunately, calculated values for the forces, moments and internal stresses were not reported.

Raftopoulos and Qassem published a method for calculation of the stresses in a bone including the three-dimensional curvature and composite nature of the femur [18]. Similar to the approach of Carter and Vasu [19], Raftopoulos and Qassem transformed the composite beam into a homogeneous beam by changing the cross-sectional geometry (area and moments of inertia) of the stiffer material. Thereafter, the composite beam could be treated as a normal beam in which the stress and strain calculations were based on Hooke's law. Calculated values for the internal loads or stresses were not reported.

Salathe et al. [20] applied beam theory to determine the stress and deformation of long bones. Cross-sectional properties were calculated for the fifth metatarsal along the curved centroid line of the bone. Assuming a distributed load along the bone long axis, the internal forces and moments were calculated. Using the axial, bending, torsion and shear components and assuming elastic properties, the stresses and deformations were computed. Due to the differences in anatomy and size that exist between the femoral and the metatarsal bones, the results, particular in relation to the load equilibrium, may not be directly transferable.

The mechanical loading of bones are the focal point of many research studies (for example on stress distribution, bone-implant interface analysis, fatigue test of implants). Therefore, it is imperative that an investigation into the load conditions of a long bone be conducted before other detailed analyses are performed. Methods capable of determining the internal loads of a femur have previously been reported [18, 20]. In addition to beam theory, finite element analysis and experimental testing techniques may be used to quantify femoral loads. In the next section, a comparison of these three methods will be provided.

1.2. Beam theory, finite element calculation and experiment

Few investigators have addressed the issue of modelling accuracy. Scholten (1975) compared the results of two-dimensional and three-dimensional finite element models with beam theory and found good agreement (factor < 1.5) between these methods within the femoral shaft region. Piziali et al. [21] evaluated the differences associated with several analytical and finite element modelling approaches. A precise and well-defined comparison of finite element methods and experimental findings has been performed by Rohlmann [22, 23]. He found comparable tendencies in stress pattern, but poor absolute agreement (factor up to 2) between strain gauge measurements taken from a loaded cadaver femur and finite element calculations for the same bone (Fig. 1). Huiskes [24] has measured the principle stresses at various levels on the femur using strain gauge techniques and compared them with those calculated from beam theory. Three force and three moment components were individually applied to the femoral head centre and the strains were measured. His comparison showed good agreement in the stress distribution using a transversely isotropic material. However, differences in the stress magnitude of up to a factor of two were reported between beam theory and the strain gauge technique. According to Huiskes, differences in the shear stress between experimental, classical axisymmetric, and finite element calculations were caused by inhomogeneties in the material used in the experiment. His goal was to perform a direct comparison of experimental and calculated surface strain data. As a result, muscle forces were not included in the model and the bone was rigidly fixed at the distal end. In conclusion the stresses found in this investigation may not be compared to those experienced during physiological loading.

Finite element modelling is a computational method, which is commonly used to analyse the stress and strain distribution in biomechanical models (Fig. 1). Femoral models, with and without endoprosthesis, have been developed by many researchers (e.g. [23, 24, 25]). A complete review of the

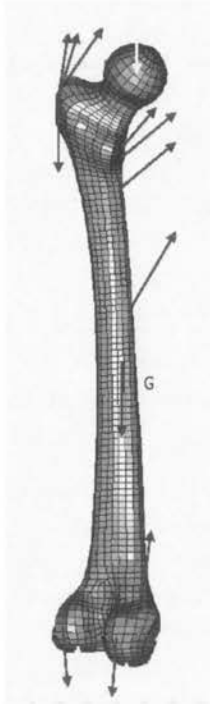

FIGURE 1. Finite element model of a femur under complex musculo-skeletal loading conditions.

literature associated with the application of this method to femoral modelling is beyond the scope of this paper. However, the establishment of load equilibrium was so far not the subject of discussion in finite element modelling of the femur. The degree to which the various muscle forces acting on the femur influence the internal stress and strain distribution has not been thoroughly investigated. In addition, many of the problems described by Huiskes et al. [24] remain unsolved: inhomogeneous material properties, which obey Hooke's law, must be considered in the application of a continuum approach. Similar difficulties exist when performing stress and strain computations with beam theory. When strictly comparing the force and moment data no differences exist among the finite element, beam theory and experimental investigations. If material property uncertainties are excluded from an analytical approach, only the forces have to be known to approximate the *in vivo* load conditions.

In summary, only a few investigators have considered muscle activity as an important influence on the load conditions in the femur [10, 13, 14, 17, 20]. Of this group of researchers, no one has attempted to use more than three to four muscle groups [10, 14, 26, 27]. There is, however, a general consensus that the muscle forces tend to reduce the loads acting within the bone. However, many of these works are restricted to analyses in two-dimensions. Furthermore, only one position from a movement or even a single activity, i.e. single leg stance, was considered as being representative for long bone loading. The criteria for performing analysis of the loads in this particular position (worst case in all six load components?) is seldom fully explained. There are no publications known to the authors that attempt to determine the internal loads in the femur resulting from a complete motion cycle or a group of daily activities.

From the reviewed literature, it can be concluded that methods to quantify the load condition in the femur are currently available. None of the reviewed publications, however, considered all forces acting on the femur. Consequently, the equilibrium of the forces and moments acting on this bone has not been fully examined. To properly construct the free body diagram associated with equilibrium, the forces and anatomy involved must be known a priori. Knowledge regarding the muscle, ligament and joint contact forces, acting on the bone is sparse. Mathematical models represent an alternative that may be used to estimate the forces in soft tissue structures and at contact surfaces. For any mathematical approach, the anatomy must be known to allow a realistic reconstruction of the load situation to enable muscle and contact force computation.

1.3. Influencing factors

The load state of a bone is defined by a combination of forces, in which the muscles attached to the femur, the joint contact forces at the hip and knee, the forces exerted by other soft tissue structures, and the forces due to

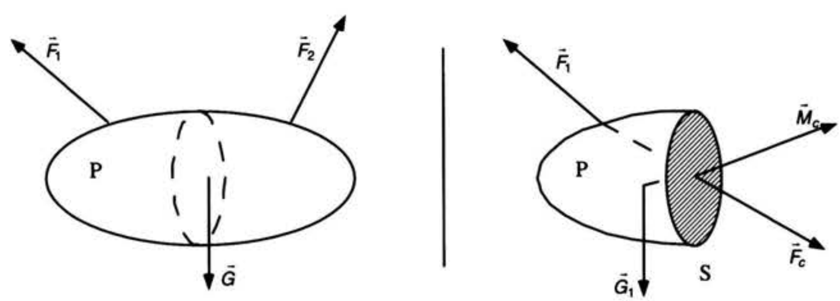

FIGURE 2. Free body diagram of a rigid body P held in equilibrium: On the left side three force vectors maintain a static position. On the right side the internal force situation is made visible in reference to the "cutting" plane S.

acceleration must be in equilibrium (Fig. 2). To fulfil the equilibrium conditions at any instance, the parameters that describe these load vectors must be considered. First, the anatomical information relating the orientation of the forces to bone position must be determined. Second, the muscle, ligament and joint force values must be generated in such way, that for a given anatomical configuration, a certain movement can be performed. Finally, the elastic properties of the materials involved must be considered because the force orientations may change with deformation of the bone.

1.3.1. Muscle anatomy. Muscle anatomy, particularly the locations of their origins and insertions, must be known to calculate the effect of muscle activity. Multiple studies have been conducted to determine muscle attachment locations. Seireg and Arvikar [28] measured the lower-limb muscle attachment co-ordinates from anatomical textbook descriptions. Crowninshield et al. [29] and Dostal and Andrews [30] located the origin and insertion co-ordinates of the leg musculature on dry bone specimens. Brand et al. [31] marked the muscle origins and insertions on three pairs of lower limb specimens and calculated the attachment co-ordinates using bi-planar X-ray images. Pierrynowski and Morrison [32] digitised the bony landmarks on disarticulated dry bones; no information was given on how the muscle attachment locations were determined. White et al. [33] located the sites of muscle origin and insertion on six pelvises and nine lower limbs using dry bones. Muscle anatomy may be derived from magnet resonance imaging data (MRI) [34]. Even though this data is more complex and allows representation of the three dimensional path of muscles, seldom groups of specimens are included to gain representative data on anatomical variations.

1.3.2. Muscle activity. To calculate the internal loads of the lower limb, knowledge of the muscle forces is essential. Approaches to quantify the relation between electromyographic signals and muscle forces have been mentioned in the previous section. Analytical methods represent yet another method that can be used to determine muscle forces. By implementation of the inverse dynamic approach, the ground reaction force of a subject can be recorded, for example, during a gait cycle [35]. If the relative motion of the body segments is known, the resultant joint forces and moments can be calculated. Due to the large number of muscles which cross the joints of the lower extremity, the use of multiple muscle force combinations may produce an appropriate resultant load situation.

From a mechanical standpoint, an infinite number of muscle force combinations allow to perform the joint movement. Such a mechanical situation is described as indeterminate since infinite solutions may lead to a single joint motion. To overcome this difficulty, two approaches have been developed. In the first approach, the number of unknowns and the number of equations are equated. Paul [36] and Ghista et al. [17] reduce the unknown forces by grouping muscles, until the equations form a determinate system.

In the second approach, the indeterminate problem is solved by applying a method that is called "optimisation". According to a specified criterion (optimisation criteria or cost function) the most suitable set of solutions is selected out of an infinite number of mechanically possible solutions. Because the problem of musculo-skeletal loading has an infinite number of possible solutions to describe the muscle force distribution, a meaningful optimisation criterion must be applied to obtain a reasonable solution. Again, two basic methods may be employed; linear and non-linear cost functions. Examples of linear optimisation are criteria such as minimizing the sum of forces [37, 38], the sum of moments [39] or a combination of these criteria [28, 40]. In general, muscle force solutions have improved with the introduction of physiologically based constraints. In contrast to methods reported earlier, these more physiologic constrains could guarantee that muscles which were reported in-active from electromyographic measurements were excluded from the calculation. Physiological based criteria include, for example, minimising the summation of stress [41, 42], the work [43], the fatigue [44], the effort [38] or the mechanico-chemical energy [45]. The above mentioned criteria can also be applied in a non-linear form, e.g. the sum of forces squared or the sum of stresses squared is to be minimized [42, 46]. In contrast to linear cost functions, non-linear functions allow to simulate a synergistic effect of muscle recruitment. While linear criteria lead commonly to a large focus in few muscles, non-linear criteria can lead to synergistic activity and tend to have more muscles with reduced muscle forces included. Multiple studies have been performed to compare the linear and non-linear approaches as well as additional optimisation criteria [47, 48, 49, 50]. In summary, calculations from the above studies indicated that the muscle force distribution was more dependent on the exact determination of the joint angles and the resultant joint moments (a factor of 3 to 4) than on the use of a particular optimisation criterion (a factor of 2 according to [45]).

The cost functions applied by previous investigators failed to include any of the effects associated with the load and motion history. The use of dynamic optimisation methods enables information about the load and movement history to be incorporated into the model [51,52]. Thunnissen found that factors, such as the force-length relationship and the physiological cross-sectional area, must be exactly determined to develop a correlation with electromyographic activities [53].

It can be conclusively stated that multiple optimisation criteria have been found to produce a correlation to electromyographic data (EMG). However, a general solution to the mechanically indeterminate problem has yet to be developed. A more complex approach which applies an EMG-based optimisation criterion could aid in the calculation of realistic muscle force data in the lower extremity [54]. An optimisation approach used in conjunction with instrumented devices [27, 55] may allow one to determine the governing equations for various muscle activities.

1.3.3. Other soft tissue structures. In addition to the forces exerted on the bone by the muscles, the force contributions of the ligaments and fascial structures must also be considered. Forces that are exerted by these soft tissue structures have a direct contribution to the state of load equilibrium. *In vivo* knee ligament forces have been reported by France et al. [56]. The joint stabilizing effects of the knee ligaments have also been investigated by a number of researchers [57, 58, 59, 60].

The anatomy of the fasciae have been previously described [61]. Investigations in regard to the magnitude and direction of the fascial forces are, however, unknown to the author. The fascial structures of the thigh combine to form three large compartments. The pressure differences that exist among these compartments directly contribute to the shear forces produced by the muscles and ligaments. Compartment pressures have been measured by previous investigators and considered of minor importance for the physiological loading of bones [62, 63, 64].

1.3.4. Femoral deformation. Multiple studies have addressed the strain distribution within the femur (usually in finite element calculations) and on the bone surface (strain gauge measurements). Combinations of both methods have been used to determine modelling accuracy. Rarely the deformation of the bone under physiological loads has been investigated. This presents another issue due to changes in the muscle force orientation during the loading process. The use of beam theory or finite element techniques would allow to calculate the model deformation. Only if the femoral deformation can be considered as small the anatomical assumptions made for an unloaded femur are equivalent to those of a femur in a loaded state (straight muscle line, etc.).

2. Validation of musculoskeletal load analysis

Measurement of loads acting in the thigh may be performed *in vitro* or *in vivo*. The use of *in vivo* measurements often reflects the variation associated with certain activities. The results from such studies are usually subject to large standard deviations. *In vivo* experiments are, however, an essential method for the acquisition of load information. If *in vivo* conditions are not required to establish and analyse biomechanical relationships (for example ligament force to joint motion), *in vitro* experiments may be the method of choice. Such analyses are often used to verify more complex *in vivo* measurements or analytical models. As mentioned earlier, *in vitro* surface strain measurements of the femur have been recorded for various applied loads. Using the measured strain data along with its geometry and material property information, one can calculate the internal forces and moments [24].

2.1. Strain measurements on bone surfaces

In a publication by Cristofolini et al. [25], the strains measured in the proximal portion of the femur were used to show the influence of muscle groups on the loading of the bone. The experimental set-up included contributions from the m. adductor longus, m. adductor magnus, mm. glutei, mm. vasti, m. rectus femoris and m. biceps femoris. A single gait phase was used to show the influence of the muscles on the strain distribution within the proximal portion of the femur. As pointed out by Huiskes [24], strain gauge measurements can only provide surface information; material and geometric properties must be known to compute the internal loads.

Cristofolini's study reported proximal femoral strains of up to ten times higher, when the influence of muscles was considered. According to Cristofolini, the mm. glutei account for 50 to 100% of the strain changes induced by muscles within the proximal femoral third. However, evaluation of the load situation at a selected instant during gait (as used by Cristofolini) provides limited information about the internal femoral loads that occur during daily activities. If muscles are to be included in experiments or calculations, it is difficult to use "correct" muscle force values. Cristofolini used a combination of muscle forces from Patriarco et al. [65] and Crowninshield et al. [41]. Each of these data sets represents, in itself, a state of equilibrium. A combination of the forces from different data sets applied to a particular anatomical position, as performed by Cristofolini, results in a totally new load situation that may not be in equilibrium. Further, due to the manner in which the forces were calculated, the mm. glutei and m. biceps femoris have rather large force values whereas the iliotibial band has none. Other muscle force calculation methods and electromyographic measurements have reported different levels of muscle activity [32, 46, 66, 67].

The influence of the mm. abductores on femoral surface strain patterns has been extensively discussed in both experimental and finite element models [68, 69, 70]. But, as stated by Rohlmann et al. [22], the vector pair mm. abductor and hip contact force, can by no means completely represent the complex loading that occurs in the femur during normal daily activities. Further investigations to determine the role that each of the different muscle groups play in femoral loading seems to be necessary.

Although the method of *in vivo* strain measurement has been used at least since 1969 [71], limited information exists about the strains in human femora [72, 73]. The results obtained from *in vitro* experiments cannot be validated without *in vivo* knowledge. According to Cordey et al. [74], the possibility of conducting accurate *in vivo* surface strain measurements could be greatly enhanced by appropriate telemetric devices. Such devices have been previously discussed by Bergmann et al. [75]. Carter et al. [76] presented a method to determine *in vivo* strain in a canine femur. His approach used beam theory and two-dimensional finite element modelling to calculate the resultant forces and moments at a particular bone cross-section. The strains, stresses and forces of a normal canine femur were then compared to those of a fractured one. It may be, however, questioned if these results can be transferred to a human femur due to the differences in anatomy and size.

Weinans et al. (1992) measured the *in vivo* loads on goat tibiae using strain gauges. Directly after sacrificing, the strain gauges were calibrated using externally applied forces. A transformation matrix was later used to calculate the corresponding forces from the measured strains [77]. Similar to the method described earlier, the load equilibrium in the goat tibia was assumed to be quite incomparable to that of the human femur. Nevertheless, *in vivo* measurements are invaluable in determining these unknown forces.

2.2. *In vivo* measurements of tendon forces

An et al. [78] presented a method for direct *in vivo* tendon force measurement using buckle transducers. Komi described a similar method to conduct transcutaneous *in vivo* ligament force measurements of the Achilles tendon [79]. He reported tendon forces of up to 3885 N ($\approx$ 6 BW) during sprinting. *In vivo* ligament force measurements represent a viable method to verify mathematical calculations of muscle forces (optimisation, reduction, etc.) [79, 80]. Such analytical distribution type solutions, with the exception of electromyographic measurements, lack validation. The use of *in vivo* ligament force measurements as a method of verifying complex mathematical calculations raises many questions. First, ethical issues have to be considered prior to *in vivo* measurements. Second, a number of unknowns must be determined to perform muscle force calculations. In contrast, direct measurements provide actual force information that can be used to validate analytical approaches. The limitations shared with other techniques of *in vivo* measurement (few patients, selected activities) remains an issue of concern.

2.3. *In vivo* measurements of joint contact forces at the hip

Hip contact forces have also been determined using *in vivo* measurements. The first measurements were reported by Rydell et al. [81, 82]. Telemetric devices were developed by many others [83, 84, 85, 86, 87]. A review of telemetric devices was given by Bergmann et al. [88]. In addition, Bergmann et al. presented the hip joint contact forces for six patients at different walking speeds [89]. The force maxima observed in this study were between 2.9 BW (2 km/h) and 4.7 BW (6 km/h) for the endoprostheses. On one occasion, a maximum resultant force at the hip of 8.7 BW was reported during stumbling. The existence of such data provides verification for calculated hip contact forces (contact force maxima e.g. Brand et al. [90]: 4–5 BW; Crowninshield et al. [29]: 3.6–5.6 BW; Seireg and Arvikar [39]: 5.3 BW). Important for a comparison of calculated and measured forces is, however, not only the maximum value but also the force distribution along a gait cycle. During the stance phase, only a slight decrease in load existed between the first and second maximum. Bergmann found this type of hip contact force pattern for both of his patients walking at various speeds; most calculations report a so called "double-peak". Conclusively, the "double peak" pattern may be a convention rather than the real pattern occurring *in vivo*. One possible explanation for the differences existing between the experimental and computational approaches may be directly related to muscle activity. The true muscle activity pattern is measured by experimentation, while an assumed pattern of activity is used for calculation.

2.4. *In vivo* measurements of internal loads in a fractured and nailed femur

Schneider et al. [91] presented *in vivo* measurements of the internal loads on an intramedullary nail that was implanted in a thirty-three year old male. The patient suffered a comminuted midshaft fracture of his left femur that was treated with a telemetric nail, similar to the AO/ASIF universal nail. Loads within the nail were monitored in supine, sitting and standing positions. According to Michel, the implant loads decreased during partial weight bearing after the seventh post-operative week by about 50%. However, during the healing process, training of the quadriceps produced an increase in the axial force ($\approx 40\%$). The authors concluded from their measurements, that even after the fracture had completely healed, approximately 50% of the loads were transmitted through the nail. Throughout the healing process, the torsional moments remained relatively small (2–5 Nm $\approx$ 0.003–0.008 BWm). The bending moments reached values of 18–22 Nm before and 4 Nm after fracture healing. The axial force had peaks of up to 300 N and the shear forces of 60–80 N during post-operative measurement (standing position, partial weight bearing). In addition, the study showed that even in static patient positions, significant axial loads (130 N $\approx$ 0.2 BW) could be produced by isometric muscular activity.

The forces and moments measured in the nail may not completely represent the load situation that exists in a healthy individual. However, they do provide information about the relative magnitude of the loads acting within the femur. The measured forces and moments were relatively small in magnitude. Therefore, it may be necessary to validate the load information associated with other dynamic activities in addition to those investigated in this study. It may also be interesting to determine to what degree the forces and moments are transferred through the fracture fragments instead of passing through the nail.

2.5. *In vivo* measurement of muscle activity

One of the largest obstacles encountered in biomechanics concerns the ability to quantify muscle force values. Multiple approaches have been used to overcome this problem. Models describing the muscle behaviour have been introduced by Hill [92] and have undergone further developed since that time. A thorough review of muscle models was performed by Winters and Woo [93].

In vivo measurements can be made using electromyography, a method in which the electric signals are measured from stimulated muscles (Inman et al., 1952). Unfortunately, not all muscles are accessible for the application of surface or needle electrodes. Also, the errors associated with signal cross-talk from the electrodes cannot always be eliminated. Approaches for transferring electromyographic measurement data to muscle force values have been developed by a number of researchers. Van Ruijven and Weijs [94] measured the activity in the jaw muscles in cat. Hof and van den Berg [95, 96] performed measurements on the human triceps surae and various calf muscles. In addition, Herzog and ter Keurs [80], introduced a technique in which the force-length relation was first calculated and than measured *in vivo*. With this approach, the change in muscle force for a group of muscles (m. rectus femoris) may be calculated.

The literature regarding joint contact forces (e.g. Bergmann et al. [89]), femoral loads (e.g. Schneider et al. [91] and muscle (e.g. Herzog and ter Keurs [80]) as well as ligament forces (e.g. Komi [79]) is somewhat limited.

These measurements were established using a very small population performing a standard set of activities. In addition, these measurement devices were directly applied to the musculo-skeletal system. It is unknown to what degree the measurement devices themselves may have modified the internal load situation.

Therefore, the use of analytical models, which attempt to calculate the internal thigh loads, remain useful and viable approaches. In parametric studies they allow the determination of the load conditions and their influencing factors. Nevertheless, *in vivo* measurements play a key role in analytical modelling by forming the gold standard necessary to evaluate the accuracy of the various theoretical approaches.

3. Musculoskeletal loading of the lower limb

Musculo-skeletal loading plays an important role in the biological processes of fracture healing [97], bone modeling and remodeling [98], and in the primary stability of implants [68]. Nevertheless, current knowledge of musculo-skeletal loading of the lower limb is still limited. While there is strong evidence that muscles are major contributors to femoral loading [99], the actual forces occurring *in vivo* are inaccessible.

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 [39, 46, 100, 101, 102, 103]. 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* [81, 83, 90]. 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. A model of the lower limb in the sagittal plane 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 [104]. 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 [105]. 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.

3.1. Loading and load history

3.1.1. 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 [83]. 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, anthropometric data was collected to determine bone dimensions, segment masses, center of gravity positions and inertia parameters [106].

3.1.2. Inverse dynamics. Clinical gait analysis was conducted for a number of different activities of every day life (Bergmann et al., submitted). 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 (Fig. 3). 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 without hand rail support. The patients selected an average stride length of 45 cm. Three patients climbed all steps (HSR, PFL, KWR) while patient IBL climbed only the first one. All measurements were taken during climbing of 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 magnitude F and 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 and the z axis is parallel to an idealized midline of the femur [83]. An optical system (Oxford Metrics, UK), consisting of a set of six infrared cameras and 24 reflective markers attached

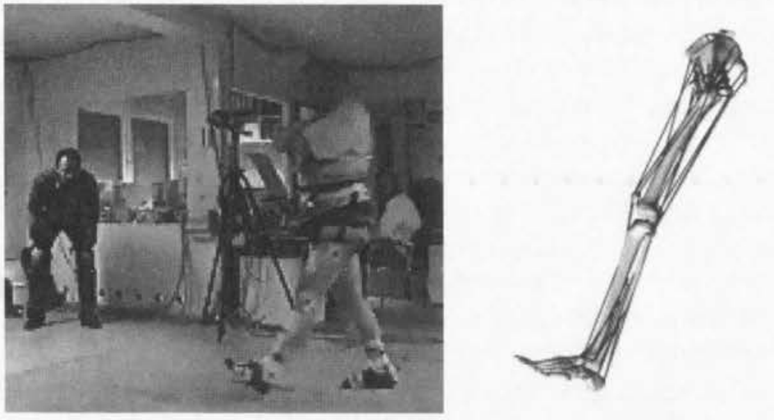

FIGURE 3. Gait analysis and musculo-skeletal analysis of lower limb loading. Hip contact, ground reactions and parameters for inverse dynamics were obtained [89] and transferred to a musculo-skeletal system for further analysis [27].

to the patients' skin, was used to determine movement of the lower limbs.

Positional information given by the skin markers and the anthropometric data were combined to derive the locations of bony landmarks. 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 [106, 107].

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 by a single patient. The relative variability was defined as the maximum difference between the peak force of a single trial and the mean peak force divided by 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 was defined as the maximum difference between the peak moment of a single trial and the mean peak moment divided by 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.

3.1.3. 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 data set 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. The bony surfaces were reconstructed from the contours [108].

Muscles were represented as straight lines spanning from origin to insertion based on descriptions from the literature [109]. 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 [109].

3.1.4. Joint kinematics. The 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 [110]. 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.

3.1.5. 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. 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 the 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.

3.1.6. Muscle and joint contact forces. Muscle force distribution was computed with a linear optimization algorithm, minimizing the sum of muscle forces [29]. Inequality constraints were imposed on the maximum muscle forces [111]. 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 [112].

A distribution of muscle forces was required to fulfill the resultant intersegmental moments at the 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.

3.1.7. 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. 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. 8.2). 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%).

3.1.8. Resultant intersegmental moments. 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 [27]. 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%)

3.1.9. Measured vs. calculated hip contact forces. Calculated hip contact forces and those measured *in vivo* were compared in all investigated trials (Fig. 4).

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. 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, 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

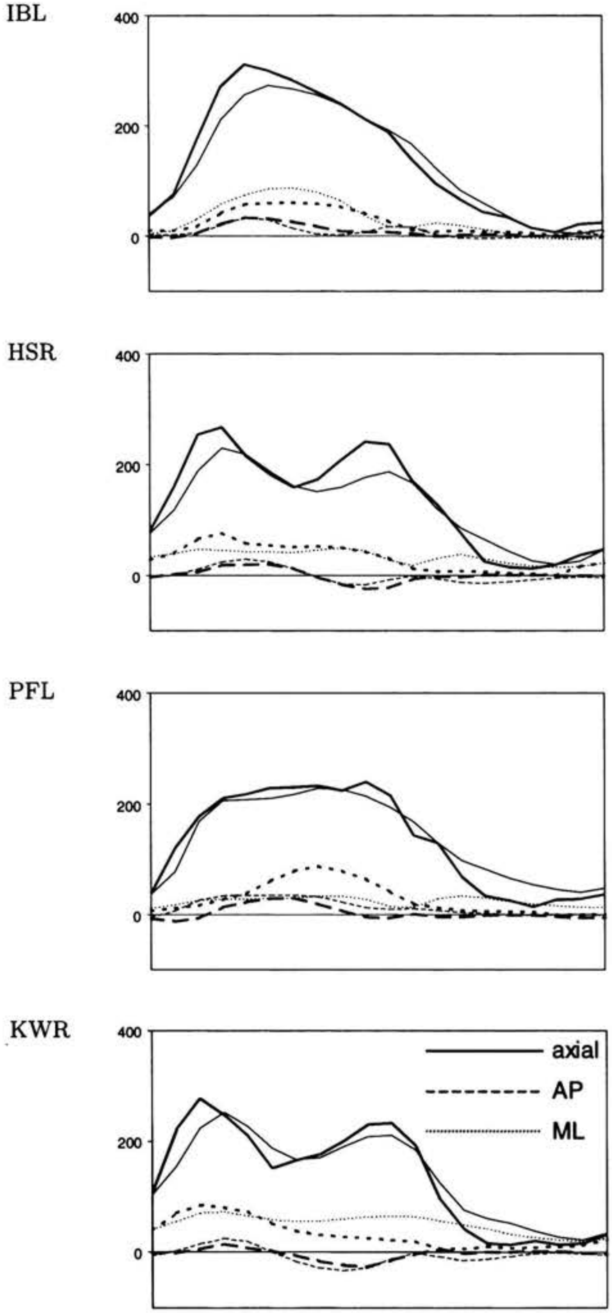

FIGURE 4. Internal loading of the bone. *In vivo* hip contact forces and calculated internal loads for four patients and two activities.

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%).

4. Summary

The aim of this study was to determine the musculo-skeletal loading conditions of the proximal femur during walking and stair climbing. While previous studies computed muscle forces [39, 40, 46, 100, 101, 103], a direct validation of the predicted loading conditions at the hip was not possible. The study of Brand et al. [55] presented measured and calculated hip contact forces in the same patient, but a 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.

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. 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 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. 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 [113].

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 [46] or the dependency of individual muscle forces on the objective function employed in the optimization calculation [114].

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.

The calculated hip contact forces and those measured *in vivo* during walking and stair climbing 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.

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