Statistical Finite Element Analysis of the Mechanical Response of the Intact Human Femur Using a Wide Range of Individual Anatomies

Mamadou T. Bah[1(✉)], Reynir Snorrason[2], and Markus O. Heller[2]

[1] University of Winchester, Winchester SO22 4NR, UK
Mamadou.Bah@winchester.ac.uk
[2] University of Southampton, Southampton SO17 1BJ, UK

Abstract. This paper attempts to obtain an improved and more physiological distribution of the applied joint and muscle forces on the intact human femur and to gain an understanding of inter-subject variability on the mechanical response. A set of 109 CT-based femur models of individual anatomies were simulated using the Finite Element method during walking. Heterogeneous material properties, physiological boundary and loading conditions were applied to each femur model to form a reference initial load configuration [1]. To correct the imbalance in the force system, an optimisation scheme was adopted that iteratively updated the locations of both muscle and joint attachments across a 5-mm radius circle centred at the initially defined node in the reference load configuration [2, 3]. Across all patients, a 28–48% reduction in the resultant reaction force magnitude measured at the femoral head was achieved. A clear gender bias was present in terms of reaction forces and strains in both the initial and optimised models. The optimisation scheme mostly affected the medial-lateral component of the reaction force. The change in the average strain was found to be highly dependent upon the percentage reduction achieved in the optimisation process. This reduction was higher for males than females and is most likely due to size differences. Body weight and bone density highly influenced reaction forces and strains. Femoral anteversion linarly increased with reaction forces; other anatomical parameters such as neck length, neck offset, and functional femoral length or CCD angle did not have a clear influence on these forces.

Keywords: Inter-subject variability · Anatomy · Femur · Walking · Finite Element · Reaction force · Strain · Optimisation · Statistical analysis

1 Introduction

With technology to enable subject-specific Finite Element (FE) analysis becoming more widely available in the field of computational biomechanics [4], the issue of variability in patient anatomy is being increasingly tackled [5, 6]. Whilst inter-individual variability also extends to the musculoskeletal loading conditions [7, 8], determination of subject specific loads might not be feasible in larger cohort studies and this variability in these internal forces is only rarely considered [9, 10]. Often, only

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typical or reference loading conditions [11] are applied to the long bones which may differ in their anatomy from those for which the loading conditions were initially derived. As a result of such a geometric mismatch, however, the force system applied to the bone may no longer be in balance. Under such conditions, the specific choice of displacement constraints to remove rigid body motion in the FE analysis can have a substantial influence on the deformation pattern and the strains in the bone [1, 12]. It is further conceivable that artefacts introduced by such unbalanced force systems mask or at least skew the true effect that variations in the anatomy can have on the strain distribution.

Previous work on the loading conditions in the femur has indeed shown that non-physiological boundary conditions can lead to substantial increase of the forces at the hip joint, reaching values of up to 4,107 N (4.7 BW) and 5,271 N (6 BW) during walking and stair climbing, respectively [1], and thus considerably exceeding previously reported in vivo data [13]. Therefore, it is important to maintain both force and moment equilibrium when conducting FE analyses. Although variation in soft tissues anatomy is thought to exceed variability in bone morphology [12], the variability of muscle anatomy could be actively exploited to re-balance the force system when a set of loading and boundary conditions is to be mapped from a reference bone to a femur with different geometry.

The objective of the present work is therefore to compare variations in the reaction forces and strain distribution in the human femur across a larger sample of individual anatomies and bone stiffnesses and using different boundary conditions. To establish reference conditions, FE analyses were first performed using a typical reference load configuration of joint and muscle forces and physiological displacement boundary constraints. Further analyses then used an optimised set of muscle and joint forces where the location of muscle and ligament attachments on the femur were varied within a range consistent with inter-individual soft-tissue anatomy variability to minimise reaction forces. For both analysis conditions, it is sought to explore the influence of key parameters of femoral morphology as well as bone mechanical properties on the hip reaction forces and the strain distribution across the sample.

2 Materials and Methods

A convenience set of 109 segmented 3D femur bone models obtained from CT scans of 75 male and 34 female subjects aged between 43 and 106 years (average age: 64.6 ± 19.7 years) formed the basis for the current study [5]. Each femur model was segmented using ScanIP software (Simpleware Ltd., UK), followed by the application of a mesh registration and morphing technique. This enabled the description of each femur model with the same number of 65031 nodes and 304638 tetrahedral elements and a unique element connectivity matrix using a previously published image-to-mesh generation process and bone density assignment [5].

2.1 Reference Load Configuration

Displacement boundary conditions were applied as described by Speirs et al. [1], see Fig. 1a. The forces (Fx, Fy, Fz) applied at the hip joint contact (P0), abductor and tensor fascia latae (P1) and vastus lateralis (P2) of the reference subject with body weight 110 kg are: (611.54 N, 311.42 N, −2524.17 N), (−713.22, −30.86, 901.89 N) and (−713.22, −159 N, −1028.29 N), respectively. Peak hip joint contact forces and eight muscle forces during walking [11] were scaled to reflect each other patient's estimated body weight. The muscle force components, defined with respect to the coordinate system defined in Fig. 1a were applied to the FE models by distributing the load over a number of nodes. First, the location of each muscle point from the reference model was mapped onto each remaining femur model to identify the corresponding node and its surrounding elements. The force components were then distributed over all of the nodes (approximately 10) associated with these selected elements. This process was employed to emulate physiological conditions where soft tissue attachments are distributed over an area but also to minimise local artefacts in the strain distribution.

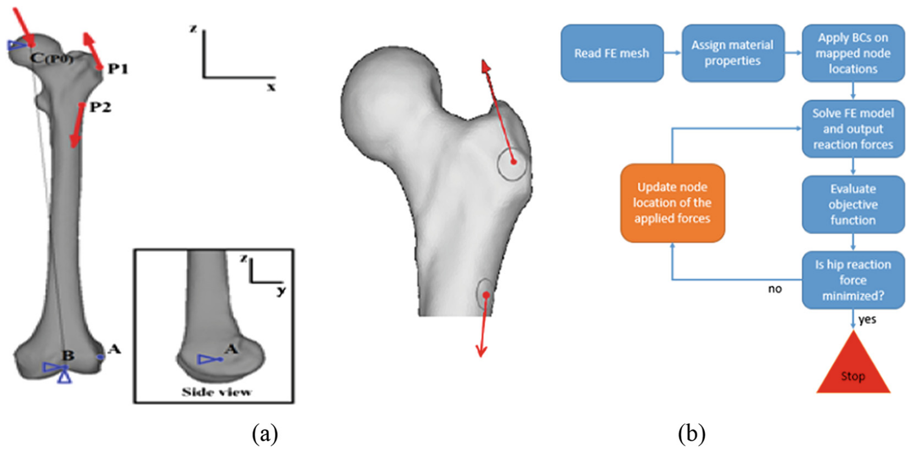


Fig. 1. The boundary conditions used for the FE models (a); the search domain and a flowchart of the adopted optimization scheme; reduced search domain where the abductor force is applied to the femur (b).

2.2 Optimised Load Configuration: Minimising Reaction Forces

A value of 10 mm for the typical variability of the location of muscle and ligament attachments [2, 3] defined the extent of the search domain to optimise soft tissue attachments. Here, a circular area with a radius of 5 mm centered at the initially defined node locations defined the permissible search area (Fig. 1b). For each femur model considered, the optimisation process involved a single objective function, i.e. minimising the reaction force on the constrained node located at the femoral head.

The design variables were the nodes at which two force vectors were applied. Each force vector can be applied to a finite number of nodes within the previously described search area. The first femur model for example led to 672 possible combinations for the two applied muscle forces.

2.3 Comparison Between Reference and Optimised Load Configuration

All FE models were solved using ANSYS Mechanical APDL (Ansys Inc., USA). Post-processing of the strain distributions focussed on the reaction forces at the constrained nodes, as well as the percentage of bone volume within specific strain limits (0–0.3%, 0.3–0.5%, 0.5–0.7% and >0.7%), and the maximum and average strain. To check for statistical differences between females and males in terms of reaction forces, two-tailed t-tests based on independent samples were performed. Finally, the individual effects of patient anatomical parameters and bone densities on reaction forces and strains were assessed and linear regression analyses were performed.

3 Results

3.1 Reference Load Configuration

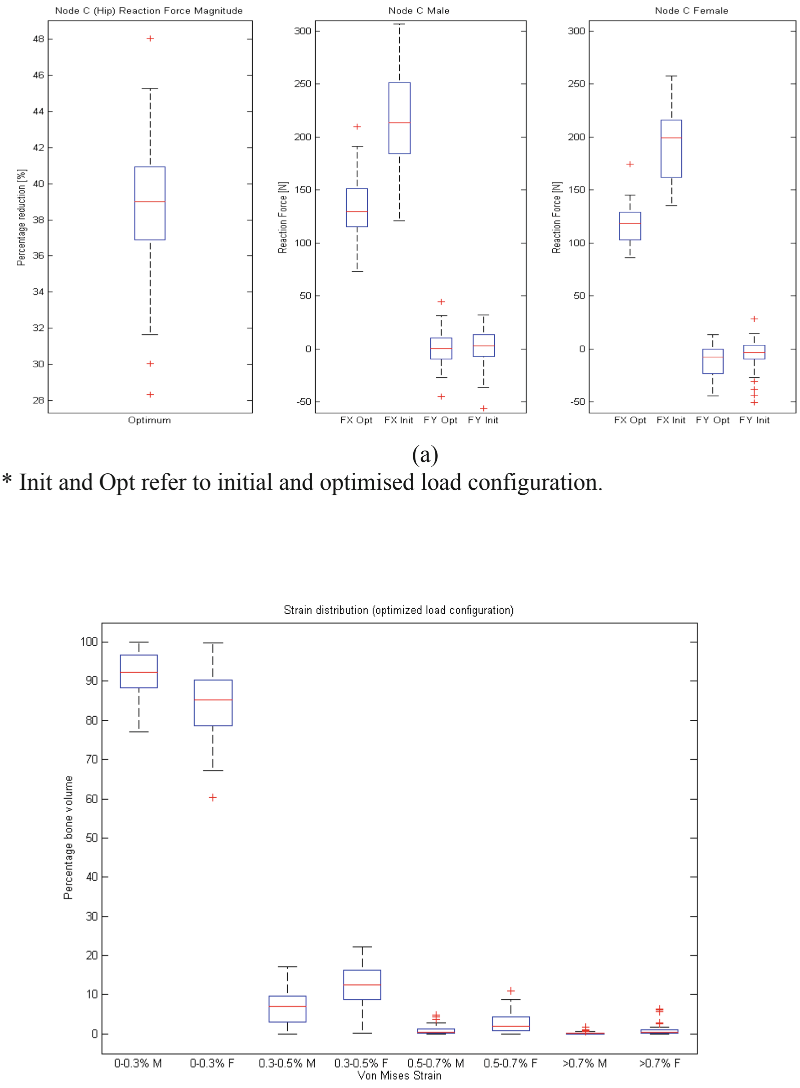
Reaction Forces. For the initial load configuration, reaction forces as high as 3000 N were recorded at the nodes where the displacement constraints were applied. The median reaction force at the distal lateral epicondyle tended to be higher for males than for females although results for females showed a wider spread of data with a maximum value of 220 N compared to 150 N for males. At the knee-joint centre, the spread of reaction forces was in general wider for males than females; the median values were also higher for males with wider 25–75% percentile ranges. At the femoral head, the median reaction force was higher in males (215 N) than females (190 N); in the medial-lateral axis the 25–75% percentile ranges of reaction forces extended up to values of 340 N and 260 N for males and females, respectively. Regression analysis showed a strong relationship between the magnitude of the reaction forces and subject weight with high coefficients of determination R^2: 1 for the Z-component of reaction force at knee-centre FZ and 0.9 for the X-component of the reaction force at head centre FX, respectively.

The analysis of the relationship between anatomical parameters and reaction forces revealed that the anteversion angle had more substantial effect on their magnitude with coefficients of determination of 0.46 and 0.3 for the posteriorly directed reaction force (FY) at the distal lateral condyle and knee centre, respectively. T-tests revealed a clear difference between males and females in terms of reaction forces ($p < 0.05$). A clear linearity was also found in the data, indicating a normal distribution.

Strain Distribution. Under the reference loading conditions, a clear difference in the strain distribution was found between females and males. In males, approximately 95% of the femur experiences strains lower than 0.3% compared to 90% in the females, who demonstrated increased average and maximum strains. No clear relationship was observed between anatomical parameters and strain distribution. However, a linear relationship was found between bone mean density and computed strains.

3.2 Optimised Load Configuration

Reaction Forces. Across the entire population, the reduction in the resultant reaction force magnitude measured at the head centre varied from 28% to 48% (Fig. 2a). This implies that varying the nodes at which soft tissue forces were applied reduced the



(a)

* Init and Opt refer to initial and optimised load configuration.

(b)

Fig. 2. Percentage reduction in hip reaction force at the head centre (a) and strain distribution in the optimised load configuration (b).

imbalance in the force system as measured by the reaction forces (Fig. 1), but could not achieve a completely balanced set of loading conditions. This Figure also revealed that the optimisation mostly affected the medial-lateral component of the reaction force FX in both females and males rather than the anterior-posterior component FY.

Strain Distribution. Similar to the reference loading configuration, a clear difference in the strain distribution was found between females and males (Fig. 2b). In males, approximately 92% of the femur experiences strains lower than 0.3% compared to 85% in the females. When the bone average strain is analysed, similar trends are observed for both females and males. Figure 3 shows an interesting relationship between the percentage difference in average strain and the percentage reduction in the hip reaction force.

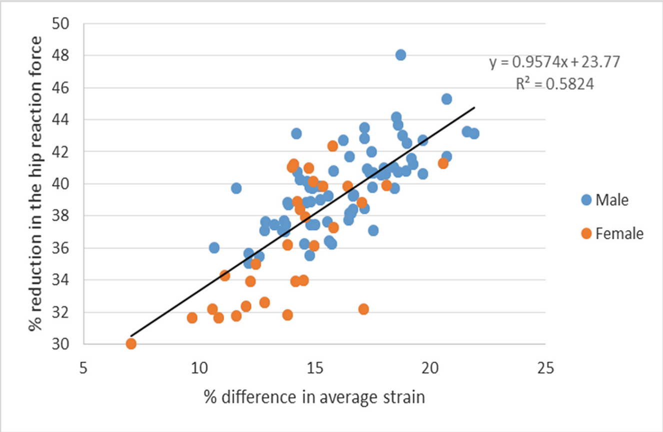


Fig. 3. Relationship between percentage reduction in hip reaction force and bone average strain.

It is demonstrated that the change in the average strain is highly dependent upon the percentage reduction achieved in the optimisation process. A higher percentage reduction in the optimisation results in a higher percentage difference in the average strain. This reduction was higher for males than females and is most likely due to size differences. Finally, when analysing the individual effects of anatomical parameters (anteversion angle, neck length, head medial offset and femoral functional length) on strain distribution, no clear trends were found.

4 Discussions

The objective of the study was to compare variations in reaction forces and strain distribution using a wide range of individual anatomies and different boundary conditions. Note that the same force vectors were applied to the femur models in both the initial and optimised load configurations. When analysing the femur with the highest reduction in reaction force, it was found that the neutral axis along the femoral shaft shifted laterally. The increased bending in the femur imposed larger strains at the surface of the cortical bone in the femoral shaft [12, 14]. This behaviour can be related to force imbalance [1, 15] and reaction forces should be reduced further in order to obtain a physiologically more accurate representation of the femur loading.

In terms of generated strains, the optimal load configuration resulted in an increase in the average strain in both male and female subjects. A decreased amount of bone volume was found within the first strain interval, 0–0.3%, after the optimisation process in both groups (Fig. 2b). A strong relationship was found between the achieved reduction in reaction forces and bone average strain (Fig. 3). Male subjects showed in general a greater reduction in reaction forces and subsequent strain distribution.

No evident effects of anatomical parameters were found except for the anteversion angle. Reaction forces are expected to depend on the geometry of the femur but not on bone elastic modulus or density. Out of the 352 possible node combinations in the optimisation, three subjects were identified (Table 1). Subject040 produced the largest percentage reduction of the reaction force (48%). Subject060 and Subject008 were found to produce the median and smallest percentage reduction, respectively.

Table 1. Anatomy data and reduction of hip reaction force reduction of seven selected subjects.

Femur ID	40	60	2	107	8	31
Gender	M	M	F	M	F	M
Age	53	79	76	50	35	55
Weight [kg]	83	47	52	133	60	72
Anteversion [°]	9.1	21.2	14.2	0.3	17.5	−0.5
CCD angle [°]	130	121	130	133.8	115.4	136.2
Neck length [mm]	55.3	54.4	47.1	47.4	47.1	48.4
Functional femoral length [mm]	411.6	386	357.7	471.1	392.7	448.2
Mean element density [g/cm^3]	0.47	0.53	0.47	0.64	0.62	0.51
Hip reaction force reduction	48%	39%	34%	38%	28%	40%

Figure 4 displays the distribution of both displacement and strain before and after optimisation for these three subjects, the ID being labelled on the undeformed femur shape. In the optimised load configuration, larger deflections can be seen; the femur is clearly under increased bending and torsion which results in even greater strain in the cortical bone on the femoral shaft. This could be expected as the reaction force on the hip was reduced significantly in the optimised load configuration. This also suggests that there is still a free moment present in the mechanical system that affects the balance. The neutral axis has also shifted laterally. Both the deflection and strain values are not far from the physiological standards of [1, 14].

The smallest female (Subject002) and tallest male (Subject107) subjects were scrutinised further. Deflections are larger in Subject107 (who had a weight of 133 kg) due to the larger applied forces. These larger deflections however, lose their physiological relevance according to the literature [14]. The reason for this is that the force and constraints models used in this study are developed for a reference femur geometry of a relatively normal size, i.e. not taking into account the extremities found within the population. Although larger deflections are observed in Subject107, strains appear to be higher in Subject002 due to the lower bone density of the subject.

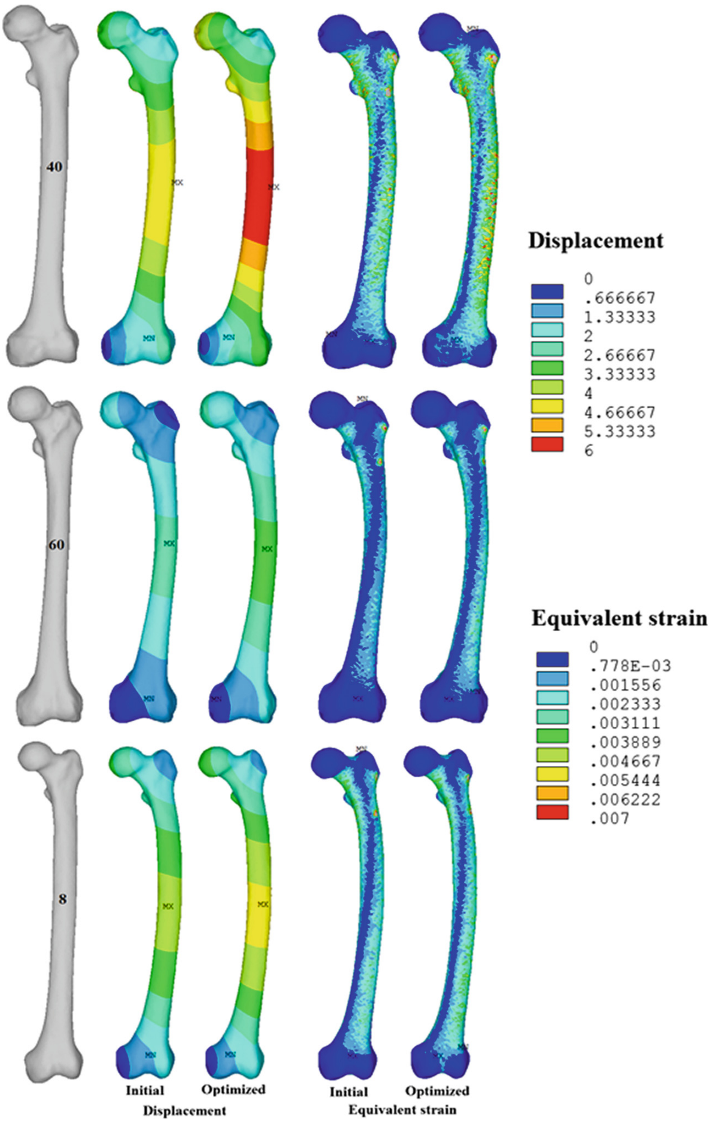


Fig. 4. Displacement (mm) and equivalent strain distribution. in three selected patients. Left grey femur shows the undeformed shape together with the subject ID number.

Subjects with the smallest (Subject008) and largest (Subject031) CCD angle were also compared, see Table 1 for more details. The mean CCD angle in males and females is 126 ± 5° and 125 ± 5°, respectively. FE results showed that Subject031 generated lower strains and smaller deflections than Subject008. This must be explained by the difference in femur anatomy, i.e. he CCD angle. For a smaller CCD angle, the force vector imposes bending in the femoral neck which results in an increased bending and torsion in the overall geometry. Increased bending and torsion leads to higher strain on the surface of the femur and larger displacements. For a larger CCD angle, the femoral neck is under compression instead of bending when the hip contact force is applied to the femoral head. This will decrease the bending and torsion in the overall geometry and result in lower strain and smaller deflections within the femur. For this specific comparison of the extremities in the CCD angle it is clear that the anatomy has an effect on the displacement and strain distribution.

In summary, a further reduction in reaction forces can be achieved by enlarging the search domain but this would lead to a design space that is physiologically not justifiable [2, 3]. The location of the knee-joint constraint could also be modified but this could impose larger bending on the femoral shaft and as a result cause higher strain and larger deflections. The knee-joint constraint modification might have influenced this behaviour but is unlikely to be causing it alone. Therefore, the location of the knee-joint constraint could be added as a design variable in the optimisation, allowing a change along the medial axis or within a spherical volume around the initially assumed knee centre node. Also, due to the uncertainty in the magnitude of applied forces, the design space could be extended further by allowing these forces to vary within specific limits to enable a shift in the balance of the force system and but also a reduced reaction force on the hip.

Conflict of Interest. All authors have no conflict of interest.

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