Investigating the mechanical response of paediatric bone under bending and torsion using finite element analysis

Zainab Altai1,2 · Marco Viceconti1,2 · Amaka C. Offiah1,3 · Xinshan Li1,2

Received: 7 September 2017 / Accepted: 13 February 2018 / Published online: 10 March 2018
© The Author(s) 2018. This article is an open access publication

Abstract

Fractures of bone account 25% of all paediatric injuries (Cooper et al. in J Bone Miner Res 19:1976–1981, 2004. https://doi.org/10.1359/JBMR.040902). These can be broadly categorised into accidental or inflicted injuries. The current clinical approach to distinguish between these two is based on the clinician’s judgment, which can be subjective. Furthermore, there is a lack of studies on paediatric bone to provide evidence-based information on bone strength, mainly due to the difficulties of obtaining paediatric bone samples. There is a need to investigate the behaviour of children’s bones under external loading. Such data will critically enhance our understanding of injury tolerance of paediatric bones under various loading conditions, related to injuries, such as bending and torsional loads. The aim of this study is therefore to investigate the response of paediatric femora under two types of loading conditions, bending and torsion, using a CT-based finite element approach, and to determine a relationship between bone strength and age/body mass of the child. Thirty post-mortem CT scans of children aged between 0 and 3 years old were used in this study. Two different boundary conditions were defined to represent four-point bending and pure torsional loads. The principal strain criterion was used to estimate the failure moment for both loading conditions. The results showed that failure moment of the bone increases with the age and mass of the child. The predicted failure moment for bending, external and internal torsions were 0.8–27.9, 1.0–31.4 and 1.0–30.7 Nm, respectively. To the authors’ knowledge, this is the first report on infant bone strength in relation to age/mass using models developed from modern medical images. This technology may in future help advance the design of child, car restrain system, and more accurate computer models of children.

Keywords Paediatric long bone · Finite element analysis · Femur strength · Injury tolerance

1 Introduction

Bone fractures are common childhood injuries which have been estimated to account for 25% of all paediatric injuries (Cooper et al. 2004). Recent evidence suggests that childhood musculoskeletal injuries may be related to higher risk of osteoarthritis later on in life (Antony et al. 2016). Children aged 3 years and younger have limited ability to protect themselves. This combined with limited communication skills, makes it difficult to diagnose inflicted injuries. In fact, the majority of fractures due to inflicted injury occur in children younger than 2 years old (Carty 1997; Loder et al. 2006). This is a big social problem with serious consequences for affected children and their families (Jayakumar et al. 2010). Unfortunately, there has been little quantitative data on bone development or strength in this age range. Previous studies were limited by sample size and available techniques to investigate the mechanical response of paediatric bone. Most of these studies focused on evaluating the strength of bone through mechanical testing (e.g. bending), which is destructive. There is also a complete lack of information on infant’s bone strength between new born and 1-year-old. This age group is particularly challenging to characterise due to the rapid growth phase (the mass of a typical neonate more than doubles in the first year of life), accompanied by marked changes in anatomy and function. Therefore, research in bone strength for very young children (0–3 years old) is highly necessary...
desirable using the latest engineering techniques. This quanti-
tative data can be used to enhance our understanding of
the injury tolerance of young bones under different loading
conditions, which are frequently presented in both acciden-
tal and inflicted injuries. This can then be compared with
the force resultant from the physical events described by the
parents/carer, combined with a dynamic model in the future.
Such data will also aid child dummy designs and the design
of car restraint systems and create more accurate computer
models of car crash tests in the future.

Fracture type is dictated by the direction and magnitude
of the applied load. Long bones may fail in tension, compres-
sion, bending or torsion, and under a combination of all or
some of these loading mechanisms. In case of an injury, the
bone will experience a combination of loads due to structural
asymmetry as well as uneven load transfer depending on the
injury scenario. The latter is more complex and requires the
construction of a full-body dynamic model to quantify the
amount of force acting on the femur for each injury scenario.
This could be investigated in the future using open source
child models such as PIPER (http://www.piper-project.eu/),
but is beyond the scope of this study. The former can be
addressed to some extent by separating these loading con-
ditions and testing each of them independently in order to
quantify a range of failure load, which is the approach taken
in this study. For example, a transverse fracture (characterised
by a fracture line perpendicular to the long axis of the bone)
may occur as a result of tensile loading and bending, whereas
a torque applied to the femur may lead to spiral fracture
(Gitajn and Rodriguez 2011; Pierce et al. 2004; Turner and
Burr 1993). Therefore, by investigating the bending and tor-
sional strength of paediatric femur, the results of this study
will provide an indicative failure range under these loads,
either separately or combined.

Previous studies have frequently used impact or quasi-
static loading on the bone in order to mimic these loading
conditions in a well-controlled experimental set-up. To date,
only a few studies have investigated fracture tolerance of
immature human bones using an experimental approach
(Forman et al. 2012; Miltner and Kallieris 1989; Ouyang
et al. 2003). Miltner and Kallieris (1989) conducted three-
point bending tests, using quasi-static loading at a rate of
50 mm/min on 28 cadaveric lower limbs (bones with sur-
rounding soft tissues). Samples were taken from children
aged 1 day to 6 years old. The reported fracture moment
ranged from 7.05 to 109.5 Nm. In 2003, Ouyang et al. per-
formed three-point bending tests under quasi-static loading
on eleven pairs of cadaveric humeri, radii, ulnae, femora,
tibiae, fibulae of children aged 2–12 years old. Bones were
freshly isolated by removing the surrounding soft tissues.
The maximum bending force reported for the femora, tibi-
ea and fibulae was in the range of 15.7–72.6, 11.3–51.3
and 2.4–20.8 Nm, respectively. The most recent experimen-
tal study on paediatric bone strength was reported by Forman
et al. (2012), in which three-point bending tests were con-
ducted on 23 cadaveric femora aged between 1 and 57 years,
using a loading rate of 1.5 m/s. They reported a bending
fracture moment ranging from 61.4 (children) to 675 Nm
(adult).

With recent developments in advanced medical imaging
and in silico technologies, non-invasive techniques such as
computed tomography (CT)-based finite element (FE) anal-
ysis have been extensively applied to study bone strength
in adults, particularly in the application of fall prediction
(Grassi et al. 2012; Lotz et al. 1991; Lotz and Hayes 1990).
The method has been fully validated against experimental
data collected from adult bones (Mccalden et al. 1997). These
studies were built on a positive correlation between bone den-
sity and modulus of elasticity. This relationship has recently
been confirmed in children in a study conducted by Öhman
et al. (2011), these authors suggested, through an experimen-
tal study on human bone tissues, that the correlation between
ash density, both strength and stiffness found in adults can be
extended and applied to children’s bones. Based on this evi-
dence, a previous study conducted by Li et al. (2015) reported
the development of a framework to model paediatric long
bones and showed initial bending analysis of 15 femora (0–
3 years old) using a CT-based FE approach (Taddei et al. 2004;
Schileo et al. 2008).

During the same period, there were a few other FE stud-
ies on paediatric bones, with one on infants and the others on
older children (6 years and above). Yadav et al. (2017) investi-
gated the effect of muscle groups’ activation on the growth of
femur using MRI-based FE models of femora for three chil-
dren (aged 6, 7 and 11 years). Both Meng et al. (2017) and
Angadi et al. 2015 investigated the accidental injury of older
children (e.g. pedestrian accidents). The former used scaled
models, generated from adult data, while the latter used pae-
diatric bone models made of simplified composite materials.
Scaled-down geometries from adults are not representative
of children as they have very different anatomy. Furthermore,
it is evident from the literature that there is an absence of
personalised anatomical or material data on infants and very
young children during the rapid growth phase, when they are
most vulnerable to inflicted injuries.

Therefore, the primary aim of this study is to quantify
the bone strengths (under bending and torsion) for 30 chil-
dren aged 0–3 years old using a CT-based FE approach. The
personalised mechanical properties will be estimated from
CT attenuation. A preliminary relationship between bone
strength and age/mass will be determined from the simula-
tion results. The information reported here will inform future
FE studies on infant and paediatric bones. These models may
also be used to create surrogate models of very young chil-
dren in the future, and combined with dynamic models to
simulate different injury scenarios.
Table 1  Demographics for the current cohort, consisting of 30 cases

| Case no. | Gender | Age (weeks) | Body mass (kg) | Height (cm) | Femur length (cm) | X-section area (cm²) | Peak modulus of elasticity (GPa) |
|---------|--------|-------------|----------------|-------------|-------------------|----------------------|----------------------------------|
| 1       | M      | 0           | 3              | 51          | 7.96              | 0.33                 | 15.67                            |
| 2       | F      | 0           | 3              | 51          | 7.49              | 0.46                 | 17.98                            |
| 3       | F      | 0           | 3              | 54          | 8.33              | 0.32                 | 18.78                            |
| 4       | F      | 1           | 3              | 56          | 8.67              | 0.58                 | 19.05                            |
| 5       | F      | 1           | 4              | 57          | 8.50              | 0.36                 | 18.88                            |
| 6       | F      | 2           | 2              | 47          | 7.74              | 0.23                 | 18.58                            |
| 7       | F      | 2           | 4              | 60          | 8.46              | 0.41                 | 18.60                            |
| 8       | M      | 2           | 4              | 55          | 8.05              | 0.39                 | 18.88                            |
| 9       | M      | 3           | 3              | 53          | 8.53              | 0.40                 | 19.91                            |
| 10      | F      | 4           | 3              | 53          | 8.44              | 0.26                 | 20.42                            |
| 11      | F      | 4           | 4              | 59          | 8.75              | 0.34                 | 22.57                            |
| 12      | M      | 7           | 4              | 55          | 9.05              | 0.51                 | 16.21                            |
| 13      | F      | 8           | 4              | 51          | 8.67              | 0.37                 | 14.09                            |
| 14      | M      | 9           | 5              | 63          | 10.10             | 0.48                 | 21.58                            |
| 15      | M      | 10          | 8              | 68          | 10.50             | 0.76                 | 16.60                            |
| 16      | M      | 11          | 6              | 58          | 9.64              | 0.49                 | 17.84                            |
| 17      | F      | 12          | 6              | 63          | 10.81             | 0.64                 | 15.43                            |
| 18      | F      | 12          | 6              | 67          | 11.14             | 0.61                 | 16.44                            |
| 19      | M      | 12          | 7              | 66          | 9.99              | 0.62                 | 16.14                            |
| 20      | F      | 12          | 6              | 63          | 10.79             | 0.77                 | 20.31                            |
| 21      | M      | 14          | 7              | 62          | 10.62             | 0.64                 | 17.39                            |
| 22      | M      | 14          | 5              | 60          | 9.60              | 0.39                 | 18.64                            |
| 23      | M      | 16          | 4              | 60          | 9.63              | 0.42                 | 16.91                            |
| 24      | F      | 16          | 6              | 65          | 10.93             | 0.66                 | 17.63                            |
| 25      | M      | 24          | 7              | 69          | 11.85             | 0.72                 | 17.29                            |
| 26      | M      | 40          | 7              | 66          | 11.00             | 0.76                 | 13.10                            |
| 27      | M      | 48 (1 year) | 13             | 83          | 15.26             | 1.26                 | 17.25                            |
| 28      | M      | 48 (1 year) | 11             | 79          | 15.17             | 1.03                 | 16.91                            |
| 29      | F      | 96 (2 years)| 13             | 92          | 18.45             | 1.39                 | 18.48                            |
| 30      | F      | 144 (3 years)| 18            | 103         | 22.41             | 2.09                 | 19.09                            |

Femur length was estimated from the CT scans representing the distance between the proximal and the distal ossification centres. The cross-sectional area of each femur was estimated through a best-fitting ellipse. The last column is the peak modulus of elasticity, estimated from the measured Hounsfield Units of the CT scans.

2 Material and methods

2.1 Finite element model generation

Post-mortem CT scans of 30 children performed at the Sheffield Children’s Hospital were used for this study. This data set has been expanded since previously reported by Li et al. (2015). Table 1 shows the geometry of the femur (length and cross-sectional area), and the material properties (modulus of elasticity) in the current cohort.

The right femur of each child was segmented using ITK-Snap (ITK, http://www.itksnap.org). The segmented bone surface was meshed with 10-node tetrahedral elements in ICEM CFD 15.0 (Ansys INC., PA, USA). Material properties were mapped from the CT scans using a well-established material-mapping procedure (Bonemat v3, Rizzoli Institute) (Schileo et al. 2008). The peak elastic modulus value in the cohort ranged between 14 and 22.5GPa. Detailed methodology is described in Li et al. (2015).

2.2 Reference system

Due to the asymmetry of the bone, a perfect alignment of the coordinate system with the long axis of the shaft would be difficult to achieve. The previous reference system reported in Li et al. (2015) ensured a good alignment in the distal half of the
Reference system used to align the finite element model for bending and torsion (anterior view); a The original reference system reported in Li et al. (2015); the x-axis passes through the long axis of the distal half of the femoral shaft but misaligns in the proximal half of the shaft. b The improved reference system where the x-axis is now defined as a line passing through the centroids of two cross-sectional areas at 25% and 75% of total femoral length. Z-axis is perpendicular to the x-y plane and points anteriorly (out of the page) following the right-hand rule.

Fig. 1 Reference system used to align the finite element model for bending and torsion (anterior view); a The original reference system reported in Li et al. (2015); the x-axis passes through the long axis of the distal half of the femoral shaft but misaligns in the proximal half of the shaft. b The improved reference system used in this study.

Fig. 2 Schematic of the bending and torsion loading conditions; a Four-point bending: two equal forces applied to the femoral shaft, adapted from Li et al. (2015); and b Torsion: moment around the x-axis was applied at the distal end while the rotational movement of the proximal end was fixed in this direction. Regions of interest for both bending and torsion are highlighted in the red boxes.

2.3 Boundary conditions

In order to investigate the effect of bending and torsion, two different boundary conditions were defined. Four-point bending was simulated by applying two equal forces to the femoral shaft in the y direction where the span of the loading is equal to half span of the supports (see Fig. 2a). Various orientations around the longitudinal axis of the shaft were then analysed with a 10° increment (35 increments in total). Only the femoral shaft (mineralised portion of the bone) was used for this simulation, which represented approximately 50% of the total length of the femur (Li et al. 2015). This was because the proximal and the distal ends of the paediatric femur were largely composed of materials in the transition to fully mineralised bone. The contribution to bending strength from non-mineralised bone would be much lower compared with the mineralised region. Furthermore, under the current boundary condition, this part of the bone would appear redundant in the simulation and was removed from the analysis.

For torsion, the full length of the femur was used in order to set up an adequate boundary condition. Two pilot nodes were added to the finite element model to define the axis of rotation along the femoral shaft. The pilot nodes were defined at the proximal and distal ends of the femur along the x-axis. The nodes were created at a distance equal to half of the total length between the centroids (at 25% and 75% of total length) (see Fig. 2b). A multi-point constraints (MPC) method was used to connect the distal and proximal nodes of the femur with the pilot nodes. The linear constrained equation related all the degrees of freedom (translation and rotation in x, y, and z directions) of nodes at both ends to the pilot nodes. The distal pilot node was fixed along the longitudinal translation (x-axis) and all other degrees of freedom were free. The proximal pilot node was constrained in y and z directions, and was not allowed to rotate in the direction of the applied moment. A torsional moment was applied through the distal pilot node with two different rotational directions representing internal and external torsion of the child’s leg.
investigating the mechanical response of paediatric bone under bending and torsion using...
Regression equations:
\[ M = a \times w + b \]
\[ M = A \times w^2 + B \times w + C \]
where \( M \) is the moment to failure, \( w \) is the mass or age, parameters \( a, b, A, B, C \) are shown below

| Age | Linear regression | Quadratic regression | Mass | Linear regression | Quadratic regression |
|-----|------------------|----------------------|------|------------------|---------------------|
| Bending | \( a=0.165 \) | \( b=0.920 \) | \( A=0.001 \) | \( B=0.073 \) | \( C=1.721 \) | \( a=1.370 \) | \( b=-3.999 \) |
| External Rotation | \( a=0.173 \) | \( b=-1.109 \) | \( A=0.001 \) | \( B=0.085 \) | \( C=1.875 \) | \( a=1.454 \) | \( b=-4.160 \) |

Fig. 4 Predicted moment to failure (Nm) plotted against the age and mass for bending and torsion; two possible regressions were indicated: linear and quadratic. The outlier is indicated with a cross. The torsion results shown are for external rotation; however, a similar trend was observed for internal rotation. The \( p \) value of all regressions are \(< 0.001\)

cohort. Consequently, body mass appeared to be a better indicator for bone strength, compared with age. The results were also suggestive of a faster increase in bone strength beyond 1 year of age (this needs to be confirmed with further cases). It is well known that mechanical factors are strongly associated with bone growth. The increase in magnitude and frequency of loading, as well as the increase in mobility in the early years, would strongly encourage bone growth in order to adapt to the evolving load. This is reflected by the increase in femur length and cross-sectional area (as shown in Table 1), accompanied by continuous mineralisation of bone tissues.

The predicted moment to failure of the current cohort, which represents bone strength, increased steadily with age and mass. As shown in Figs. 4 and 5, the predicted trend was in agreement with previous experimental studies, where a positive relation between fracture moment and age was also reported (Forman et al. 2012; Ouyang et al. 2003). The range of fracture moment predicted by the current FE model under four-point bending was 0.85–27.9 Nm. Although this range is comparable to that reported in Ouyang et al. (2003) and Forman et al. (2012) (see Fig. 6), it should be noted that the bending moments reported in these two studies were measured under three-point bending tests, which were likely to predict higher strength than four-point bending. Three cases from Ouyang et al. (2003) were within the age range of the current study, at 2, 2.5 and 3 years, respectively. The reported
Investigating the mechanical response of paediatric bone under bending and torsion using...  

Bending

Regression equations:
\[ M = a \times w + b \]
\[ M = A \times w^2 + B \times w + C \]
where \( M \) is the moment to failure, \( w \) is the mass or age, parameters \( a, b, A, B, C \) are shown below

| Age     | Linear regression | Quadratic regression |
|---------|-------------------|----------------------|
| Bending | \( a = 0.165 \)   | \( A = 0.0004 \)     |
|         | \( b = 0.920 \)   | \( B = 0.107 \)      |
|         | \( C = 1.526 \)   | \( C = 3.999 \)      |
| Mass    | \( a = 1.370 \)   | \( A = -0.015 \)     |
|         | \( b = 1.761 \)   | \( B = 0.762 \)      |
|         | \( C = 1.715 \)   | \( C = -0.758 \)     |

Fig. 5  Moment to failure (Nm) plotted against the age and mass under bending and torsion for age ranged between 0 and 6 months old to illustrate the trend in the infant age range. The \( p \) value of all regressions are \(< 0.001\)

fracture moments were 29.6, 24.3 and 39.6 Nm, respectively. Three cases reported in Forman et al. (2012) fell within our age range, one at 1.33 years old and another two at 2 years old. The fracture moments of these femora were 61.4, 61.7 and 65.5 Nm, respectively. The reason for the comparably higher fracture moment predicted by Forman’s study could be due to the much faster loading rate used (1.5 m/s compared with 0.008 m/s in Ouyang’s study). Furthermore, Ouyang’s study was based on a Chinese population, whereas Forman’s study was based on a Spanish population. Consequently, ethnicity may play a role in the measured differences in bone strength. It should be noted that there was no comparable experimental data for children below 1 year old, where the majority of the current cohort sits. To the authors’ knowledge, there is only one study in the literature that reported three-point bending tests on 28 children aged between 1 day and 6 years old (Miltner and Kallieris 1989). Quasi-static loadings were used for 18 specimens while dynamic loadings were used for the remaining 10 tests. This study reported a fracture moment ranged from 7.05 Nm (6 days) to 109.5 Nm (6 years). However, the entire lower limb was used in this study including all surrounding soft tissues. Previous literature suggests that soft tissues (including skin, fat and muscles) surrounding the bone may absorb some energy during loading. Therefore, the results of Miltner’s study are not readily comparable with those reported here (Kerrigan et al. 2003)

We could not find any study in the literature that performed a torsional loading test on paediatric bones in order to com-
Due to scarcity of data, it was assumed that the high correlation between the ash density and both strength and stiffness found in Öhman et al. (2011) (4–15 years old) can be extrapolated to represent the age range in the current study (0–3 years old). The predicted failure strength was then compared against previous experimental data (Forman et al. 2012; Miltner and Kallieris 1989; Ouyang et al. 2003), which were confirmed to be within the same range. Future work will need to be conducted in order to confirm the validity of this assumption and validate the FE model. One potential approach could be through the testing of young animal bone, as reported in Cheong et al. (2017).

The current study suggests that the finite element approach, which has been widely and successfully used in adults, can be adapted and applied to children. Since getting paediatric bone samples is very difficult, using non-invasive techniques such as CT-based finite element analysis provides a valuable alternative to the investigation of paediatric bone biomechanics. In future, this technique will allow us to create surrogate models for infants and very young children, as well as obtaining more quantitative information about bone growth and strength to drastically enhance the little information currently available in the literature.

Acknowledgements This study was supported by the Higher Committee for Education Development in Iraq (HCED). The project has also received funding from the MultiSim Project (EP/K03877X/1) and from the European Commission H2020 programme through the CompBioMed Centre of Excellence (Grant N. H2020-EINFRA-2015-1-675451).

Compliance with ethical standards

Conflict of interest The authors declare that there is no conflict of interest.

Open Access This article is distributed under the terms of the Creative Commons Attribution 4.0 International License (http://creativecommons.org/licenses/by/4.0/), which permits unrestricted use, distribution, and reproduction in any medium, provided you give appropriate credit to the original author(s) and the source, provide a link to the Creative Commons license, and indicate if changes were made.

References

Angadi DS, Shepherd DET, Vadivelu R, Barrett TG (2015) Orthogonal digital radiographs - A novel template for a paediatric femur finite element model development, vol 46, p 58. https://doi.org/10.1007/978-3-319-11776-8
Antony B, Jones G, Jin X, Ding C (2016) Do early life factors affect the development of knee osteoarthritis in later life: a narrative review. Arthritis Res Ther 18:202. https://doi.org/10.1186/s13075-016-1104-0
Bayraktar HH, Morgan EF,Niebur GL, Morris GE, Wong EK, Keaveny TM (2004) Comparison of the elastic and yield properties of human femoral trabecular and cortical bone tissue. J Biomech 37:27–35. https://doi.org/10.1016/S0021-9290(03)00257-4
Investigating the mechanical response of paediatric bone under bending and torsion using...

Lotz JC, Cheal EJ, Hayes WC (1991) Fracture prediction for the proximal femur using finite element models: part II—nonlinear analysis. J Biomech Eng 113:361–365. https://doi.org/10.1115/1.2895413

McCalden RW, Mcgeough JA, Court-brown CM (1997) Age-related changes in the compressive strength of cancellous bone. J Bone Jt Surg A79:421–427

Meng Y, Pak W, Guleyupoglu B, Koya B, Gayzik FS, Untaroiu CD (2017) A finite element model of a six-year-old child for simulating pedestrian accidents. Accid Anal Prev 98:206–213. https://doi.org/10.1016/j.aap.2016.10.002

Miltner E, Kalleris D (1989) Quasi-static and dynamic bending stress of the pediatric femur for producing a femoral fracture. Zeitschrift fur Rechtsmedizin J Leg Med 102:535–544

Öhman C, Baleani M, Pani C, Taddei F, Alberghini M, Viceconti M, Manfrini M (2011) Compressive behaviour of child and adult cortical bone. Bone 49:769–776. https://doi.org/10.1016/j.bone.2011.06.035

Ouyang J, Zhu Q, Zhao W, Xu Y, Chen W, Zhong S (2003) Biomechanical character of extremity long bones in children and its significance. Chin J Clin Anat 21:620–623. https://doi.org/10.1017/CBO9781107415324.004

Pierce MC, Bertocci GE, Vogeley E, Moreland MS (2004) Evaluating long bone fractures in children: a biomechanical approach with illustrative cases. Child Abuse Negl 28:505–524. https://doi.org/10.1016/j.chiabu.2003.01.001

Schileo E, Taddei F, Malandrino A, Cristofolini L, Viceconti M (2007) Subject-specific finite element models can accurately predict strain levels in long bones. J Biomech 40:2982–2989. https://doi.org/10.1016/j.jbiomech.2007.02.010

Schileo E, Dall’Ara E, Taddei F, Malandrino A, Schotkamp T, Baleani M, Viceconti M (2008) An accurate estimation of bone density improves the accuracy of subject-specific finite element models. J Biomech 41:2483–2491. https://doi.org/10.1016/j.jbiomech.2008.05.017

Taddei F, Pancanti A, Viceconti M (2004) An improved method for the automatic mapping of computed tomography numbers onto finite element models. Med Eng Phys 26:61–69. https://doi.org/10.1016/S1350-4533(03)00138-3

Theobald PS, Qureshi A, Jones MD (2012) Biomechanical investigation into the torsional failure of immature long bone. J Clin Orthop Trauma 3:24–27. https://doi.org/10.1016/j.jcot.2012.02.001

Turner CH, Burr DB (1993) Basic biomechanical measurements of bone: a tutorial. Bone 14:595–608. https://doi.org/10.1016/0756-3282(93)90081-K

Yadav P, Shefelbine SJ, Pontén E, Gutierrez-Farewik EM (2017) Influence of muscle groups’ activation on proximal femoral growth tendency. Biomech Model Mechanobiol. https://doi.org/10.1007/s10237-017-0925-3