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Research paper

Finite Element Analysis on Pelvis With Leg Length Inequality

N. F. Othman[1], H. Y. Tan[1], K. S. Basaruddin[1]*, M. H. Mat Som[1], W. M. R. Rusli[1], A. R. Sulaiman[2]

[1]School of Mechatronic Engineering, Universiti Malaysia Perlis, Pauh Putra Campus, 02600 Arau,Perlis.
[2]Department of Orthopaedics, School of Medical Science, Universiti Sains Malaysia, 16150 Kubang Kerian, Kelantan, Malaysia
**Corresponding author E-mail: khsalleh@unimap.edu.my*

Abstract

Leg length inequality, also known as leg length discrepancy (LLD) is a condition which the left and right legs of an individual are noticeably different in length. When the level of LLD is high, such as those of 20 mm and above, it would disturb the wellbeing of an individual in terms of gait, and also causes them to experience higher stress in their pelvis compared to individual without LLD. In order to study load due to LLD had affects human bones such as the pelvic bone, finite element analysis (FEA) approach is usually used as it allows limitless attempt to investigate the stress-strain response on human bones and is far more practical than experimenting on real bones, therefore FEA was done with by using ANSYS 15.0. From the data obtained via FEA, the risk of fracture can be calculated, which gives us an insight on how would LLD affects the risk of bone fracture. In this study the effect of pelvic tilt caused by LLD has been studied, along with how loads at various LLD level affects the pelvic bone. The verdict from the study is the pelvic tilt caused by LLD amplifies the maximum stress and strain on the pelvic bone. The analysis using hip load due to LLD shows a downtrend for the maximum stress caused by the longer limb as the level of LLD increases, while the maximum stress caused by the shorter limb shows an uptrend with the increment of LLD. The maximum stress and strain observed are usually distributed around the sacroiliac joint. It is also observed that the higher the level of LLD is, the higher the maximum stress on pelvic bone will become, hence the higher the fracture risk.

Keywords: *Finite Element; Fracture Risk; Leg Length Inequality; Pelvic.*

1. Introduction

Leg length inequality, or more commonly known as leg length discrepancy (LLD), is a condition where the left and right legs of an individual are noticeably unequal in length [1]. According to Donald [2], it is estimated that as much as 70% of the world's population exhibits symptoms of LLD. A lower degree of LLD, typically below 20 mm would not affect a person much as compensational behaviour will take place [3] by adjusting their gait and overall movement kinematics to minimize the effect of LLD with little to no side effect to the individual. However, LLD with the severity of 20 mm and above will significantly affect a person's gait due to the imbalance limb, and the compensation behaviour could lead to amplified forces across a smaller joint contact area [3, 7]. The claim can be supported by Bhave et al's [8] study which discovered significantly different ground reaction force between the limbs of an individual diagnosed with LLD.

LLD is commonly compensated naturally by the human body through adjustment in the pelvis region by the change in pelvic obliquity, and when the severity of LLD is higher there's a tendency for the human body to flex the knee of the longer leg to aid in the compensation for LLD [11]. This LLD induced pelvic tilt causes the centre of gravity of an individual to be shifted from its normal position, which could disrupt the balance of stress-strain distribution and also increases the internal joint load at hip joint [3,6,9,10]. The increment in the internal joint load causes the hip joint to experience an abnormal amount of joint contact force which is higher than usual, which could lead to discomfort and even risk of injury when an individual with LLD engage in sports activity such as running, as demonstrated in Bradley's study where the amount of runners with LLD who reported hip pain is twice as much as the runners who don't have LLD [12]. Furthermore, there is a possibility of more serious injuries such as stress fracture due to the shifting centre of gravity which causes imbalanced weight bearing in individuals with LLD, according to McCaw et al's [13] study where he found that the likelihood of a person to experience stress fracture in their lower limb increases with the increment of the severity of LLD. Since pelvis is linked to lower limb, it is likely that the pelvis region is negatively affected by the increment in ground reaction force caused by LLD, and it will be one of the interests of this study to look on the relationship between LLD and fracture risk. While there is an increasing amount of study interest involving LLD, the study of the pelvis region which is also affected by LLD, particularly the hip region, does not attract as much of study interest compared to other regions in lower limb such as the knee and ankle, therefore leaving a lot of questions and concerns involving pelvis of LLD patient unexplored. Hence, this paper aims to look into this fairly unexplored area of study to help opening up a path for the science community to better understand the effects of LLD on pelvis region, focusing on its mechanical aspects, particularly on parameters such as stress-strain response around the pelvis region. The stress analysis becoming a popular alternative to analysed the stress fracture due to bone remodelling [14,15], tissue properties and loading transfer.

2. Methodology

2.1. Geometrical model of lower limb bone

In order to be able to carry out a simulation with high accuracy, an accurate 3D model of the lower body bone is needed. Such 3D model is usually obtained by doing a CT scan on an individual,

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and then the CT scan is converted into a 3D model using software, and finally, the model is processed to refine its surfaces. However, the fundamental of human lower body limb bone used for this project was obtained from a database of TurboSquid. Figure1 below shows the geometrical model used in this study.

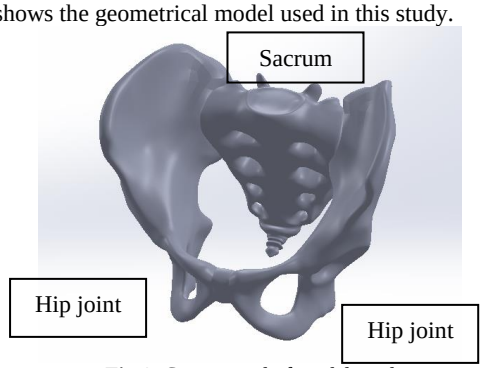


Fig 1: Geometrical of model used

2.2. Finite element model of pelvic

Nine stages of LLD will be simulated in this study, with the load values obtained through the experiment, and applied into ANSYS for simulation. The 3D model, particularly the pelvic girdle will be adjusted to simulate LLD by tilting it by a 0.216° per millimeter, a value obtained from Vink et al's [17] study.

The simulation will be divided into two type, namely simulation of LLD under varying pelvic tilt and simulation of LLD under varying hip load. For simulation of LLD under varying pelvic tilt, each stage of LLD, separated by intervals of 5 mm, will have a different degree of pelvic tilt corresponding to their level of LLD. A constant load will be applied onto the hip joints across all the levels of LLD, and the simulation will be carried out on an assumption that the subject was a healthy male with a body mass of 62.8 kg. For simulation of LLD under varying pelvic tilt, loads during standing condition and walking condition will be used, and it is assumed that both scenarios are under double leg stance condition, in which both of the legs are in contact with the ground. Typically during a quiet standing condition, the forces acting on the hip joints are around 0.5×BW [8], while the hip reaction forces typically range from 2.4×BW to 7.5×BW [19, 1]. For standing condition, the load used will be set at $F = \frac{62.8 \times 9.81 \times 1/3}{2}$ for each hip joint, whereas for a walking condition, the load used will be set at $F = \frac{62.8 \times 9.81 \times 3}{2}$ for each hip joint.

For simulation of LLD under varying hip load, the simulations will be separated into two parts as well, with one set of simulation focusing on the longer leg loads, and another set of simulation focusing on the shorter leg loads. The magnitude of the loads were obtained experimentally from the artificial LLD (by using attached insole) study previously, with the magnitude obtained at the instance of single leg stance, producing around the peak value of peak hip joint contact force[4], [5].

The material properties for the pelvic bone were assumed to be linear elastic and isotropic, therefore the materials were sets as Isotropic Elasticity found under Linear Elastic tab during the creation of the material. The material properties such as Young's Modulus and Poisson's Ratio were decided based on findings from the literature review, the selected material properties were shown in Table 1.

Table 1: Material properties for bone [12]

Structure	Young's Modulus	Poisson's Ratio
Bone	17 GPa	0.3

2.3. Boundary conditions

For the boundary condition of the finite element model, the top surface of the sacrum, at the region which it connects with the vertebrae is appointed as a support for the geometrical, as demonstrated in Figure 2 by point C. For LLD simulation under varying pelvic tilt, the forces acting on the pelvic bone are placed on the acetabular joint parallel to the vertical y-direction, as shown by point A and B in Figure 2. As for LLD simulation under varying hip load, there will be only one hip joint contact force applied at a time, either on point A or B, because LLD simulation under varying hip load is set up to be under the condition of single leg stance, hence only one side of the acetabular joint will experience joint contact force at a time. As for the contacting surfaces between the pelvic bone and sacrum, it is set as bonded to imitate the characteristic of the sacroiliac joint.

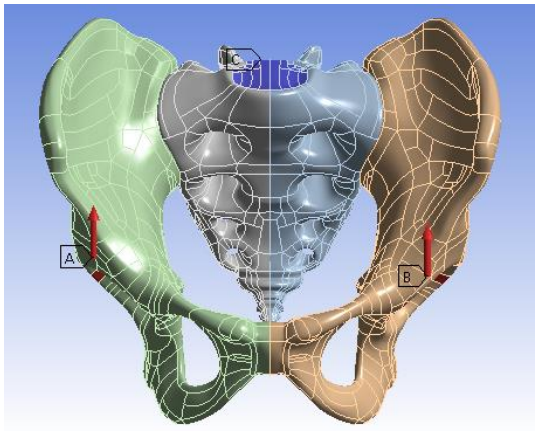


Fig 2: Points of boundary condition

2.4. Data analysis

Once all the simulation result of all nine stages of LLD for all of the conditions were obtained, all the FEA results will be compiled and compared for analysis. The analysis will be focused on the stress-strain distribution, in which the pattern and the region of focus of the stress-strain distribution on the geographical model will be analysed, in order to understand the stress-strain behaviour on pelvis.

Besides, the fracture risk on pelvis and also femur will be determined as well by calculation. The formula for bone fracture calculation is as follows:

$$\text{Fracture risk} = \text{Bone Fracture Strength}/\text{Von-Mises Stress} \quad (1)$$

The von-Mises stress will be determined from the result of FEA, and the highest value of von-Mises stress for each stage of LLD will be taken for the calculation. The fracture strength of bone will be set to 205 MPa based on Keaveny et al's [21] work.

3. Results and discussion

3.1. Stress and strain distribution

Figure 3 shows the contour of stress-strain distribution produced from by 5 mm LLD during standing condition. It was observed that there were 4 critical points focusing on the maximum stress and strain distribution, at regions around the sacroiliac joint (SIJ) where the pelvic bone and sacrum meets. The stress and strain distribution for the rest of the simulations displayed similar behaviour, with the difference being slight variation in the shape of maximum stress and strain and their respective values.

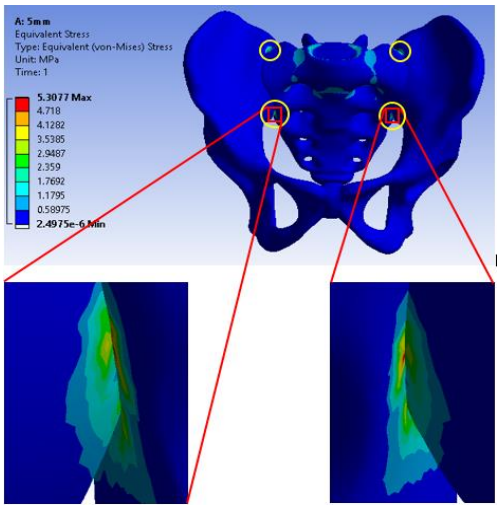
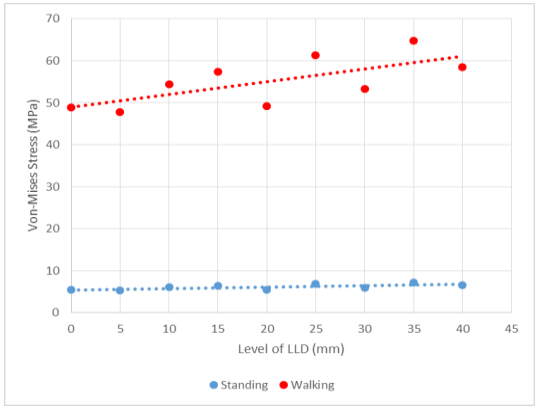


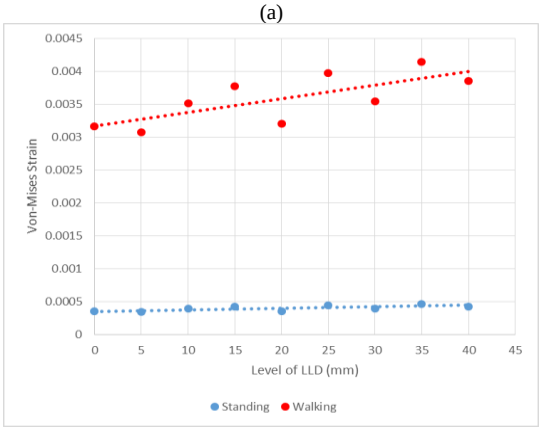
Fig 3: Contour of stress response for 5 mm LLD during standing condition

3.2. Simulation of LLD under varying pelvic tilt

The maximum stress and strain response on the pelvic bone during standing condition can be seen in Figures 4(a) and (b). It was observed that there's an uptrend with both of the stress and strain response as the level of LLD increases. This indicates that the higher the severity of LLD it is more likely to experience a higher stress and strain on the pelvic bone. The lowest maximum stress and strain experienced during standing condition were 5.3077 MPa and 0.00034299 respectively at 5 mm LLD, while the highest maximum stress and strain values were 7.1989 MPa and 0.00046076 respectively at 35 mm LLD. The disparity between the highest and lowest recorded under this series of a simulation were 1.8912 MPa for maximum von-Mises stress and 0.00011777 for maximum von-Mises strain.



(a)



(b)

Fig 4:(a) Maximum Stress Response under Standing & Walking Condition due to LLD; (b) Maximum Von-Mises Strain Response under Standing & Walking Condition due to LLD

For the simulation of LLD under varying pelvic tilt during walking condition, the pelvic bone experiences nine times as much force when compared to standing as the body experiences forces equivalent to about 1/3×BW while standing and about 3×BW while walking, the higher magnitude of force causes a higher maximum stress and strain response as well as demonstrated in Figures 4(a) and (b).

Similar to the stress-strain response under the effect of LLD during standing condition, the stress-strain response during walking show uptrend as well as the LLD level goes up, but at a much higher degree in terms of its gradient of trend line when compared to the stress-strain response during standing condition. The lowest maximum stress and strain were recorded at 47.769 MPa and 0.003081 respectively at 5 mm LLD, while the highest value of maximum stress were 64.79 MPa and 0.004147 for the highest value of maximum strain, both observed at LLD level of 35 mm. The difference between the highest and lowest recorded under this series of simulation were 17.021 MPa for maximum von-Mises stress and 0.001066 for maximum von-Mises strain.

In general, the behaviours of stress distribution is similar to the pattern of strain distribution, as in the values of maximum stress-strain reaction of walking condition is significantly higher than walking. Furthermore, the gradient of the trend line for stress-strain distribution under walking condition is noticeably higher than the gradient of the trend line for stress-strain distribution during standing condition.

3.3. Simulation of LLD under varying hip load

3.3.1. Simulation of longer limb load

Based on the results obtained from simulating LLD with focus onto the right leg, it was observed that the maximum stress reaction starts at a significantly lower than average value when the level of LLD is 0 mm which indicates no LLD, as shown in Figure 5, and then soar by a staggering 64.43% when the first stage of LLD was applied at 5 mm, afterwards it went into a downtrend as the LLD gets higher up to 40 mm. When LLD occurs, the maximum stress exhibits a downtrend as the level of LLD goes up, this indicates that the stress caused by the longer limb of an LLD patient would go down as the inequality of the legs become higher.

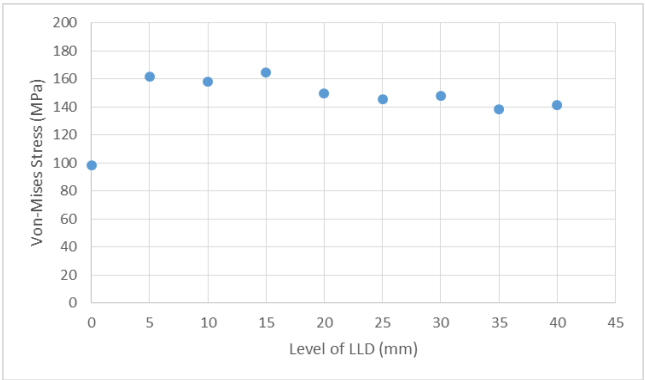


Fig 5: Maximum Von-Mises Stress Response on Longer Leg

The maximum strain distribution caused by the load on the long leg of an individual with LLD exhibits a similar behaviour with the pattern shown by the stress distribution caused by same load. As demonstrated in Figure 6, at 0 mm of LLD, the maximum value of strain was lower than average, and then moving forward to the next stage of LLD at 5 mm it increased by as much as 59.24%, and then proceeds to go into downtrend as the level of LLD increases.

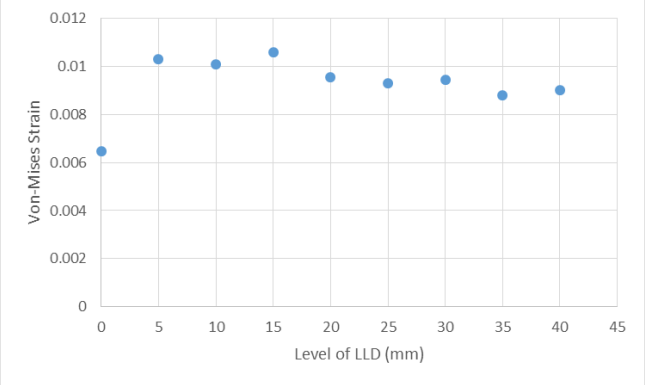


Fig 6: Maximum Strain Response on Longer Leg

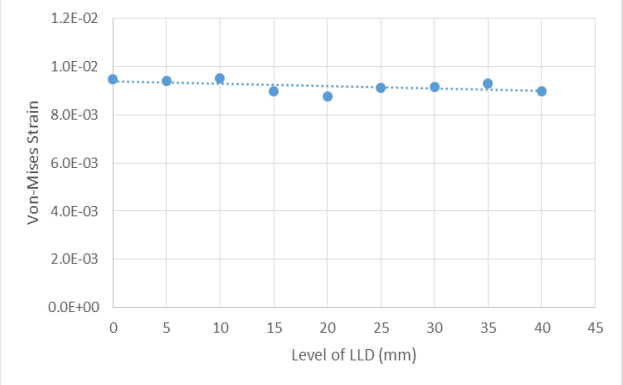


Fig 8: Maximum Strain Response on Shorter Leg

As for the maximum stress-strain distribution under the influence of longer leg load of an individual with LLD, the highest value of stress and strain were recorded, 161.51 MPa and 0.010326 respectively at 5 mm of LLD. As for the lowest value, the stress and strain went as low as 98.23 MPa and 0.0064846 respectively at 0 mm LLD, and then the next lowest values recorded would be 138.25 MPa for stress and 0.0088058 for strain, both occurred in 35 mm LLD. When only conditions with LLD were considered, the percentage difference between the highest and lowest value for maximum stress was 15.52% while the percentage difference for maximum strain was higher at 45.70%.

3.3.2. Simulation of shorter limb load

Based on the graph in Figure 7, the trend of the stress caused by the load of left leg of an individual with LLD in general rises as the level of LLD increases. This trend was the opposite of what was exhibited from the simulation done using data collected from the longer leg, which the maximum stress value drops with the increment in LLD level.

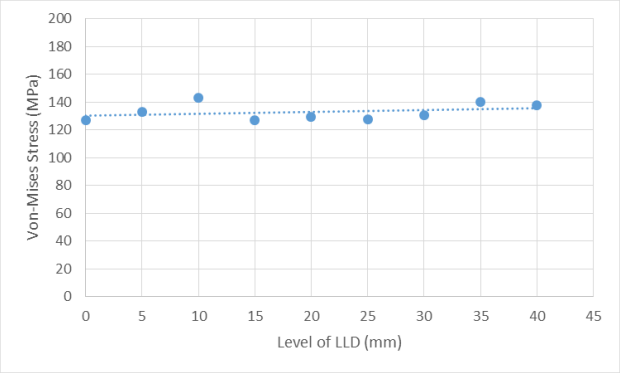


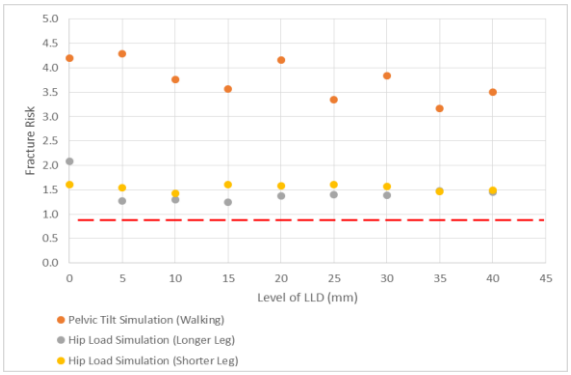
Fig 7: Maximum Stress Response on Shorter Leg

As for the maximum strain caused by the longer limb of a person with LLD, the strain experienced by the pelvic bone decreases as the length discrepancy of the legs increases, as demonstrated in Figure 8. This behaviour is different from all the previous scenarios where the general trend of maximum strain across different level of LLD exhibits similar trend with the behaviour of maximum stress.

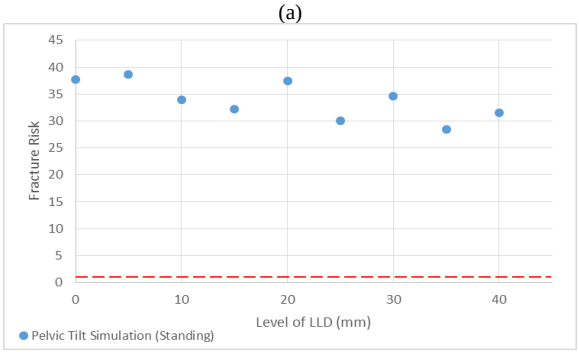
For the mechanical stress-strain distribution caused by the shorter leg of an individual with LLD, it is shown that the highest maximum stress recorded was 143.27 MPa whereas the highest maximum strain observed was 0.0095136, both results were from LLD level of 10 mm. On the other hand, the lowest maximum stress registered was 127.25 MPa at 0 mm of LLD, and then the next lowest maximum stress would be 127.31 MPa at 15 mm of LLD, which is slightly larger than that of 0 mm of LLD by only 0.06 MPa. As for the lowest maximum strain, it was recorded at the level of LLD of 20 mm with a value of 0.0087674. When only conditions with LLD are taken into factor, the percentage difference for maximum stress was 11.80%, whereas the percentage difference of maximum strain was lower at 8.16%.

3.4. Assessment for fracture risk of pelvic bone due to LLD

The data for fracture risk of each level of LLD under various simulation scenario was compiled and shown in Figures 9(a) and (b). Bone fracture is expected to happen should the value of fracture risk drops below 1, which is below the red dashed line of the graphs below. As a whole all four of the series of data exhibits an increasing risk of fracture as the level of LLD increases.



(a)



(b)

Fig 9: Risk of Fracture of each Level of LLD for Pelvic Tilt Simulation; (a) Standing; (b) Walking

For pelvic tilt simulation of both standing and walking condition, the fracture risk was far from the failure zone of below 1, especially for simulation during standing condition with the lowest value being 28.477, as shown in Figure 9(a). On the other hand, the lowest value for simulation under the walking condition is considerably high as well at 3.164 hence there should be no concern of fracture risk in the two simulated conditions. For pelvic tilt simulation, both standing and walking condition shown a percentage difference of 30.24% between the highest and lowest fracture risk.

For hip load simulation, both longer and shorter legs exhibit a relatively high fracture risk compared to those from pelvic tilt simulation. It is observed that for as the level of LLD goes up, there is a trend where the fracture risk caused by the shorter leg increases, while the fracture risk caused by the longer leg decreases. The closest values to fracture risk found were 1.243 for the longer leg and 1.431 for the shorter leg. As for the percentage difference between the highest and lowest value of fracture risk, the percentage differences of the fracture risk caused by the longer leg and the shorter leg were 17.61% and 11.77% respectively.

3.5. Discussion

The general consensus here is LLD amplifies the forces acting on the pelvic region, as the length discrepancy between the legs increases the reaction forces experienced by an individual will be higher as well, and such effect is greater when a person is in a more active state compared to when being stationary, as suggested from the results of the simulation of LLD by adjusting the angle of pelvic tilt according to each LLD level, where both standing condition and walking condition of LLD is compared.

In the simulations where the left and right leg loads are applied and simulated separately, it is found that the forces acting on the right acetabular joint affects mainly on the right part of the pelvic bone and only cause a barely noticeable stress-strain reaction on the left part of the pelvis. While for the forces acting on the left acetabular joint, it is the opposite as the focus of stress-strain distribution moves towards the left side of the pelvic bone. This means that most of the stress-strain reaction of the left and right side of the pelvic bone is mainly caused by the respective limbs as the limb from the other side has a very minor effect towards the stress-strain reaction that's not on the same side of the limb.

Moving further, it is also found that the higher stress-strain reaction are usually focused on the upper and lower regions of the SIJs, hence there are four regions that are affected with stress and strain more than the other regions of the pelvic bone. While there's stress-strain contour found on some parts of the sacrum as well, it is less notable as compared to the stress-strain reaction found on the SIJs as the contours displayed on sacrum are all of low values, making it relatively insignificant as compared to the stress-strain reaction on the SIJs.

In the hip load simulations, it is found that as the level of LLD increases, the maximum stress caused by the longer limb decreases, whereas for the shorter limb causes the maximum stress in the pelvis region to increase proportionally to the level of LLD. This indicates that for an individual with LLD, the discrepancy of the leg length causes each of the legs to induce the different amount of stress onto the pelvic bone, with the longer limb imposing lesser stress with the increment of LLD level while the shorter limb causes higher stress as the level of LLD goes up. Furthermore, when comparing the stress response between the longer limb and lower limb, the longer limb generally experiences more force hence the maximum stress built-up on the pelvic bone is higher, which leads to a slightly higher fracture risk compared to the shorter limb. The trend of longer limb experiencing greater force than the other limb is supported by findings from Muratagic et al.[22], which found an identical trend in terms of force experienced by the limbs.

Moving forward, when the stress-strain response increases in its maximum value due to having LLD, there is a concern if the higher stress caused by having LLD will increase the chance of getting a bone fracture at the pelvic region. The fracture risks are calculated based on the data obtained from the simulation, and it is shown that as the level of LLD increases, which causes the stress reaction value to increase as well, the risk of fracture also increases. Both of the pelvic tilt simulations show that generally as the level of LLD increases, the fracture risk increases as well. As for hip load simulation, it is learned that while the increment of fracture risk is proportional to LLD level, the longer leg and shorter leg have different trend for fracture risk in response to LLD level. The shorter leg causes a higher chance of fracture risk, while the longer leg has a decreasing trend of fracture risk as the level of LLD increases. While no fracture risk value goes below the value of 1, some of the values are still considerably low, which could go as low as 1.243. The pelvic tilt simulation under walking condition and hip load simulations have some disparity between their fracture risk values even though both are simulations of LLD under walking condition, the reason behind that is because pelvic tilt is done under double leg stance condition, which both legs are in contact to the ground, distributing the forces experienced by an individual into the two legs, whereas the hip load simulation was done under single leg stance condition, hence the forces are exerted and focused into a single leg, causing higher stress concentration on a region, resulting on a higher fracture risk. Depending on the gait, the hip reaction forces can range from 2.4×BW to 7.5×BW [19, 20] during walking, with double leg stance producing lower value while single leg stance producing a higher value.

4. Conclusion

Through the outcomes of this study, the objectives of the study have been successfully achieved. A 3D geometrical model of the pelvis has been developed for the purpose of conducting finite element analysis, with its material properties and suitable mesh set, along with a series of a 3D model with variation in terms of pelvic tilt. By using the 3D geometrical model developed, the mechanical stress-strain distribution caused by LLD under various conditions were simulated and the set of data was collected and studied upon. From the data of maximum stress collected, the fracture risk due to LLD has been calculated to provide an insight of how LLD affects the risk of bone fracture of the pelvis.

In conclusion, LLD generally amplifies the stress and strain acting on the pelvis, and with that, it imposes a higher risk of fracture towards the pelvic bone. Therefore, when an individual is diagnosed with LLD, which occurs in many individuals, they should take cautious towards the issue and seek for medicinal advice if the LLD of a particular individual is high, as a high level of LLD could impose health risk with the amplified stress and strain on the bones, especially near the joint area as found from the result of this study.

Acknowledgement

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