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The effect of leg length inequality on joint contact forces of lower limbs during walking

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Purpose: The aim of this study was to examine the joint contact forces (JCF) between each limb as the LLD magnitude increases during walking activity. *Methods*: Eighteen male healthy subjects volunteered to participate in the experiment. Walking gait analysis was conducted with eight different levels of insole to simulate the LLD, starting from 0 cm until 4.0 cm with 0.5 cm increment. Qualisys Track Manager System and C-motion Visual 3D biomechanical tools were used to analyse the results. Four joints (ankle, knee, hip, and pelvis) of lower limb of two legs were investigated. The increment of insoles was placed on the right leg to represent the long leg. *Results*: The results suggest that the mean contact forces for all joints in the short leg were increased as the increment level increased. On the contrary, the mean contact forces in the long leg decreased when the LLD level increased. Among these four joints, JCF in hip shows a positive increment based on the ASI value. Means that hip shows the most affected joint as the LLD level increase. *Conclusions*: The result obtained in this study might help clinicians treat patients with a structural LLD for treatment plan including surgical intervention.

Key words: *leg length inequality, leg length discrepancy, joint contact force, lower limb*

1. Introduction

Impairment between two legs caused by leg length inequality or also known as leg length discrepancy (LLD) provoke mechanical and functional changes in terms of gait pattern and posture alteration, changes in joint moments and power as well as unbalance loading transfer within each segments [1]–[3]. Small magnitude of LLD, commonly classified as "mild LLD", also caused the repetitive loading on specific joint and segment [4]. Treatments warranted for this problem are based on the level of discrepancy. Usually, corrective treatment is conducted in surgical way when the level is greater than 2.0 cm [5]. If the level below than 2.0 cm, the clinician advices to insert an insole over the patient shoes or sandal in order to reduce the detrimental effect caused by LLD [6]. It is confirmed that the overcorrecting would later develop problems on bone, cartilage, musculoskeletal limb and posture, especially during post-operative session. A surgeon carries a huge responsibility on identifying which part is to be sacrificed that will not give a damage on their patient's bone. However, functional impairment can preserve post-operatively, particularly when the outcome of the surgical procedure is technically sub-optimal, even if the operation was success.

Either short or long leg, one of these limbs would induce more forces, as compensation mechanism occurs during existent of LLD condition. The compensation mechanism involves lengthening of the short leg by flexing the long leg, so that individual with LLD can control the balance as well as contact forces

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induced on their joint pain during gait. There is such evidence that an abnormal transmitting load on both limbs may predispose to a degenerative joint disease likewise a knee, hip or lumbar arthritis [7]. A development of superolateral hip arthritis was found on the longer side [3], increase stress joint moment on the acetabulum cup of hip on the short limb [1], increase of the ground reaction forces on the long leg [8] and decrease in push-off force in long leg while increase of the weight acceptance rate on the shorter limb [9]. In terms of anatomical changes, the pelvis tilt and obliquity were being widely discussed [5], [10]–[14]. Other affected joint such as knee and shoulder, were occasionally discussed [15].

The additional effect of LLD's magnitudes with the induced loading (between short and long limb) on lower limb joints remains unclear. Numerous kinematics studies of LLD have emphasised the effect of congenital LLD [3], [16] and shoe lift [17]. Along with time, the congenital LLD had better accommodating motion pattern and joint loading as compared to an individual who is late-onset with the LLD case, like in a patient with a joint disease such as osteoarthritis. Besides, the altered gait pattern and altered joint contact forces have been related to imbalance contact forces between limbs. It is plausible that an individual with LLD tends to demonstrate gait alteration, resulting in altered contact forces, and to silence the pain of their intact limb. In the literature, there are many reading materials on the effect of LLD on the hip and pelvis joint. Effect on the knee and the ankle joint is infrequently discussed. Therefore, this study aims to evaluate the effect of LLD increment on joints contact force response in lower limb during normal walking.

2. Methods

2.1. Subjects

Eighteen normal male subjects, without any orthopaedic problems or medical history of fractures and no leg length difference were recruited in this study. Their median age was 22 ± 1 years, with an average height of 1.67 ± 0.05 m and weight of 62.8 ± 7.8 kg. All subjects were selected based on normal Body Mass Index (BMI) in the range of 19 to 23. All subjects were briefed before they signed the written informed consent forms. This study also received an approval from the ethics committee.

2.2. Gait protocols

Five motion capture cameras (Oqus Qualisys 100) with a 200 Hz sampling rate, and two force-plates (Bertec) were used for data acquisition, as shown in Fig. 1. Calibration was performed with a T-wand and L-shaped metal frame in order to minimize the noise between inter frame during data collection. The technical error for this setup within a working volume as less than 0.1 mm. Any reflection surface was covered so that only the reflective markers were captured during the experiment. Thirty-one reflective markers (20 mm in diameter) were attached on the subject's lower limb. These markers were then adhered with double-sided tape before they were added to the subject's skin. The arrangement of the markers was adopted from Horsman's method (Fig. 2). Kinematics and kinetics

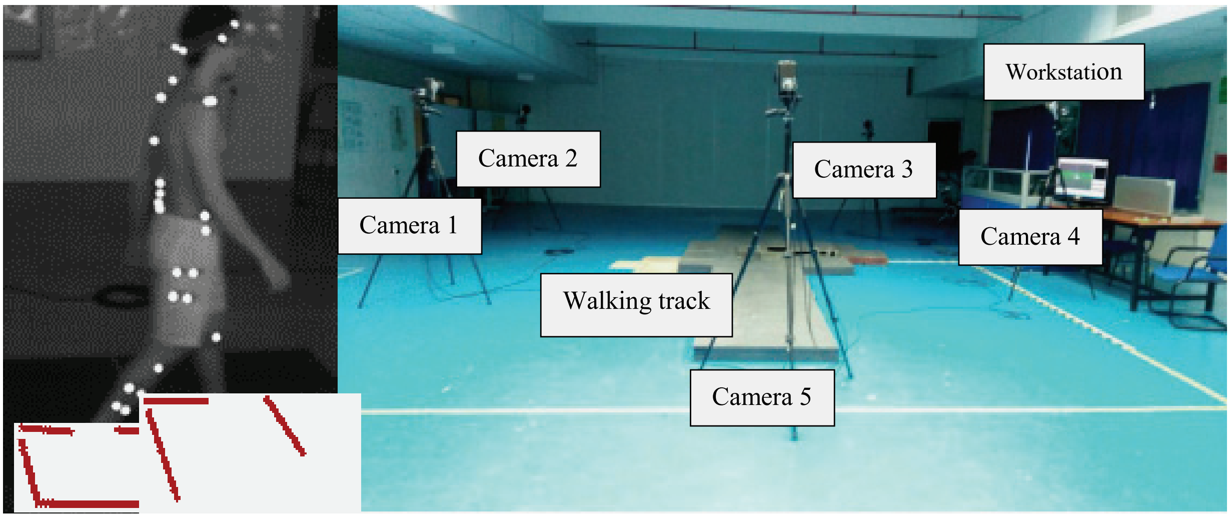
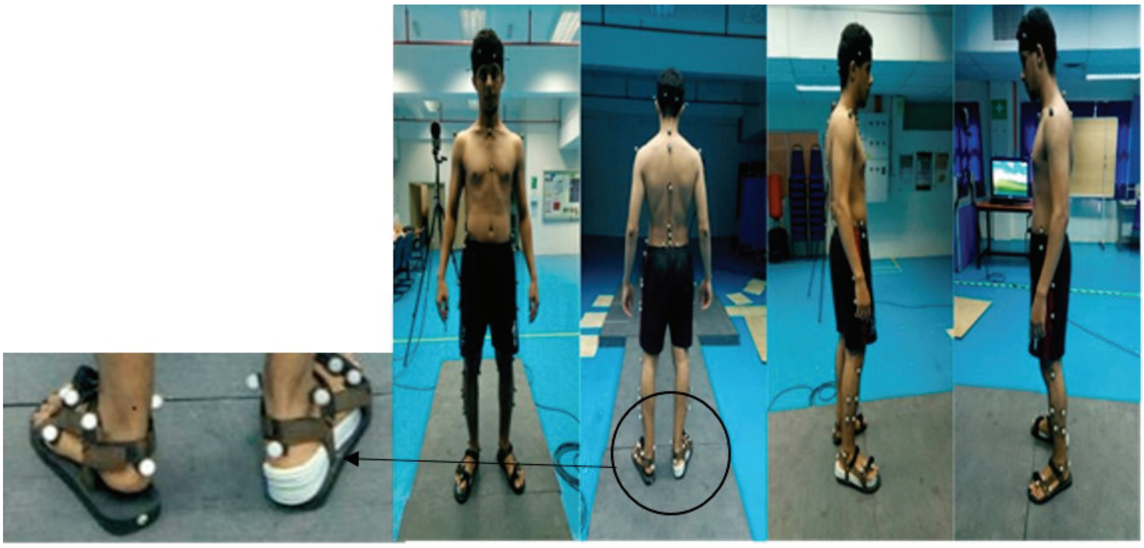


Fig. 1. Layout of motion capture system: (a) Markers captured and force plate location, (b) Arrangement of experimental equipment



(a) (b)

Fig. 2. (a) Magnified view of insole attachment, (b) Marker arrangement based on Horsman's method

parameters for lower limbs data in three-dimensional were obtained by Qualysis Track Manager (QTM) motion capture analysis system version 2.6.

All subjects were asked to walk on a 7-meter walking track, with an attached insole on the right sandal to mimic the LLD condition, at their self-selected speed (normal walking). Subject must ensure only one foot was used to step down on each force plate. Prior to the experiment, all subjects were asked to practice walking on the track with the sandal. They were asked to do this trial three to five times in order to familiarize them with the environment and to encourage them to walk naturally. Eight increment levels of insoles were used starting from 0.5 cm to 4.0 cm. The interval between each of the insole was 0.5 cm. Consequently, the data was captured three to five times, then, the best captured data was further analysed. Pre-processing data were done in QTM. Gait kinematics data for stance phase were cropped and then exported to a C3D file name format. The experimental procedures in the present study have been validated in the previous study [18] by comparing the joint contact force (JCF) responses between the simulated LLD subject and a true LLD patient.

2.3. Biomechanical evaluation

The ankle, knee, hip and pelvis contact forces were analysed using a three-dimensional linked-segment model of C-Motion Visual 3D (C-Motion Inc., Germantown, MD, USA). There are four segments defined by the markers placement: bilateral feet, shank, thigh, and the pelvis. Data in kinematics and ground reaction force was calculated based on inverse dynamics in order to obtain joint contact forces. The musculoskeletal model was developed based on an anthropometric dataset provided in the conventional gait model. The model was then scaled to reflect joint position and subject's body size.

All data were filtered using a low-pass fourth order Butterworth filtering techniques, with a cut-off frequency of 10 Hz. All joint contact forces were normalised with the subject's body weight (BW) to eliminate the influence of BW variation among each subject. Absolute symmetry index (ASI) was calculated to determine the significant differences in contact load between both limbs. A zero index represents no significant difference between both limbs, hence the gait is considered symmetrical [19]. The ASI values were rounded to the nearest integer in order to notify the differences between each limb. Then, the gait symmetry between each limb was analysed for contact forces on every segments. The ASI value was calculated based on Eq. (1) [20]:

$$\mathrm{ASI} = \frac{|X_{\text{long}} - X_{\text{short}}|}{\frac{1}{2}(X_{\text{long}} - X_{\text{short}})} \times 100\%, \qquad (1)$$

where X is the joint contact force.

2.4. Statistical analysis

Mean values with a 95% confidence intervals (CI) were measured in order to quantify the variation between the subjects. Since most of the gait data were not normally distributed, a non-parametric approach was employed in statistical analysis. Mann–Whitney *U*-test was performed to compare the significant differences between the short limb and long limb as well as between the control levels (0 cm) to the other LLD levels. A significant difference was obtained if the *P*-value is less than 0.05. All statistical analyses were performed using the IBM SPSS Statistics 20.

3. Results

The joint contact forces with respect to the increment levels of the LLD for ankle, knee, hip, and pelvis are shown in Fig. 3. The joint contact forces for short and long leg were computed to determine the significant changes when the LLD level induced. The mean value for all segments of the short leg indicates that the increasing trends occur, compared to a level of 0 cm and 4 cm.

3.1. Joint contact forces

The results show that the ankle contact forces in the short leg has a greater value than in the long leg, as shown in Fig. 3a. The pattern for 0 cm to 0.5 cm level of LLD was seen to be decreasing both for short and long leg. Then, at a level of 1 cm, this value drastically increases for both legs. It is also noted that at LLD level of 1.5 cm there is a decline in mean value for both limbs. Interestingly, the length of LLD levels of 1.5 cm until 4 cm, there is a consistent increase in value of ankle contact forces for both long and short limb magnitudes. In Fig. 3b, the mean value for knee contact forces was found to be greater in the short leg, which shows an overall of increasing trend. These results are somewhat with the long leg results, which shows an average decrease in trends. The result for the short leg demonstrates an increase in mean value starting at 0 cm prior to 1 cm level of the LLD. Abruptly, the value in our data study drops slightly through 1.5 cm and increases gradually until 3.5 cm of the LLD. After that, it shows a little decline in 4 cm discrepancy. Meanwhile, results of the long leg exhibit slight decrease on 0 cm to 0.5 cm. After that, through a level of 1 cm up to 2.5 cm, an increase in trends is noted. Interestingly, a fluctuating trend is present in LLD level of the 3 cm prior to 4 cm.

As can be seen in Fig. 3c, the greater mean contact force value is presented at the short limb rather than at the long limb of the hip segment. Both short and long legs present the same trend patterns in graph at the 0.5 cm level, the mean contact forces show a decline in value, compared to the 0 cm. Then, it is slightly increased at the level of 1 cm, whereas along level of 1.5 cm till 2.5 cm, the value on the short leg is consistent throughout. Nevertheless, the long leg exhibits a contradicting pattern which is in the declining of the mean value, on par with the length of with LLD level 1 cm to 1.5 cm. However, the values show a consistent pattern at the length of 1.5 cm up to 2.5 cm of the discrepancy, then, they gradually decline above LLD levels of 3 cm prior 4 cm.

As can be seen from the graph in Fig. 3d, an average pelvis contact forces of short leg indicates the incline of the pattern throughout all levels except for 1.5 cm. During LLD level of 2 cm, there is a gradual increase in mean value until a level of 4 cm. Yet, the trend for long leg shows inconsistency. There are also declined in mean value on 1.5 cm and then it became constant along LLD levels of up to 3 cm. There is a slight increasing trend on LLD levels of 3.5 cm, then declines again on LLD level of 4 cm.

3.2. Symmetric measurement between limbs

Based on the ASI value (Table 3) on ankle segment, during LLD level of 0 cm to 2.5 cm, the value marked no symmetrical difference (0–1%). Meanwhile, starting at the LLD level of 3 cm, there are asymmetrical percentages detected between both short and long leg length (2%). A further statistical test revealed that the significant changes occur only at LLD level of 4 cm, for asymmetric difference on both short and long limb ($P = 0.01$). However, none of other LLD level shows a significant difference ($P > 0.05$). A Mann–Whitney *U*-test (Table 1) revealed that significant difference was obtained when comparing between LLD levels and normal (0 cm). The Mann–Whitney *U*-test results showed the significant changes of LLD level (0.5–4 cm) were only detected at level of 3.5 cm and 4 cm for both the short and long leg ($P < 0.05$). Other LLD levels showed no statistical significant difference ($P > 0.05$).

Further analysis of the ASI percentages of the knee contact force shows that LLD level of 0 cm until 2.5 cm are symmetric between both the limbs (0–1%). Whilst, an asymmetry between limbs detected on LLD level of 3 cm to 4 cm (2–4%). This is in line with the statistical analysis performed (Table 1) to pinpoint the difference between the limbs, which revealed that

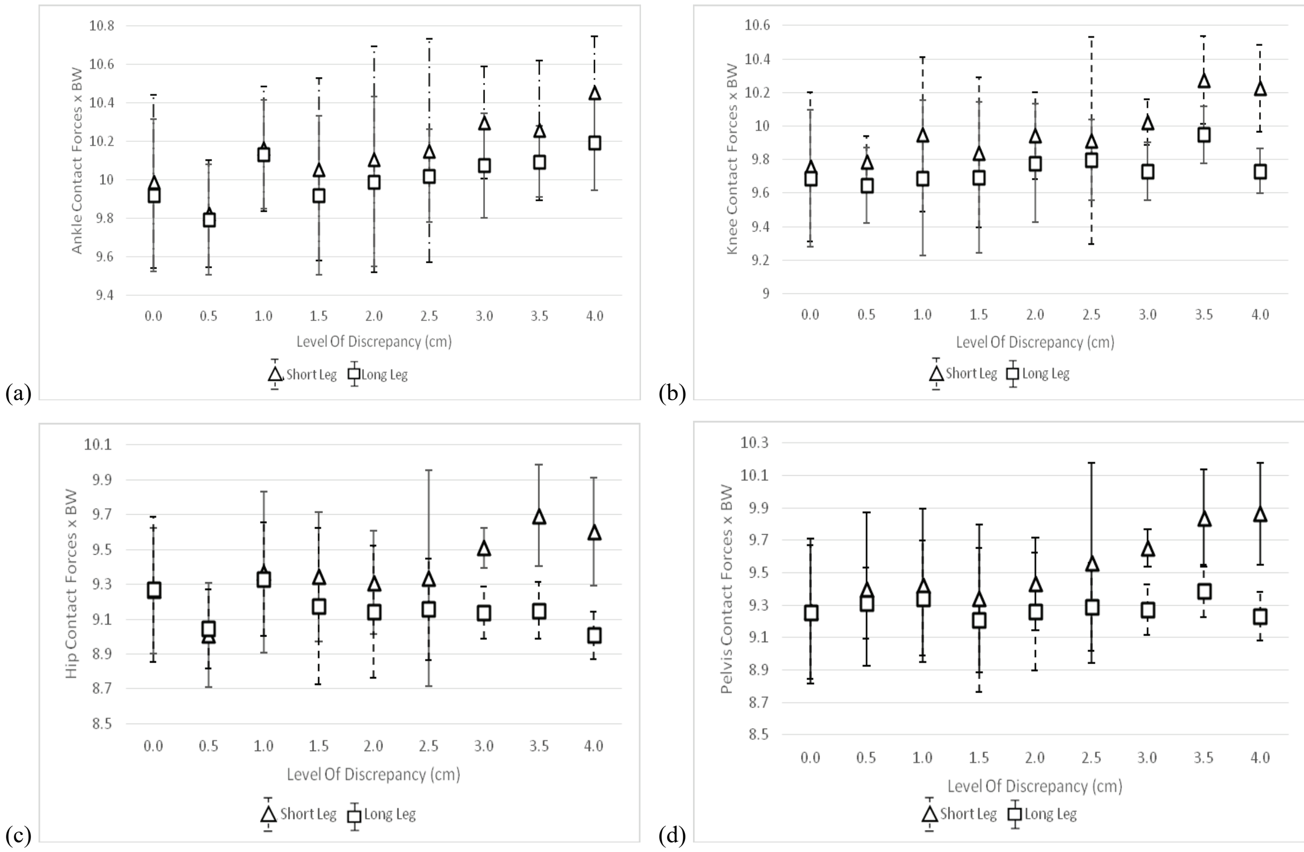


Fig. 3. Mean (±standard deviation) for lower limb segments:
(a) ankle, (b) knee, (c) hip and (d) pelvis joint contact forces based on the increment level of LLD

there are significantly differences on a level of 3.5 prior to 4 cm ($P < 0.05$). Other levels does not reveal any significant difference ($P > 0.05$). Thus, the test to identify the affected level when being compared to 0 cm, exhibits that only at the level of 4 cm had significant difference ($P = 0.01$). The rest of the results come out with no significant difference ($P > 0.05$).

Next, an analysis was performed based on hip segment. ASI percentages show that none of asymmetrical detected on a level of 0 cm to 2.5 cm (0–1%), whereas the asymmetrical between limbs presented on level 3 cm prior to 4 cm (5–6%). These results are then supported by statistical analysis done using the Mann–Whitney independent test between limbs (Table 2). There are significant difference existing on level 3.5 cm and 4 cm respectively ($P < 0.05$). The remaining levels do not match with any significant difference ($P > 0.05$). Another statistical analysis performed to spot the affected level, compared to 0 cm. It shows that only 3.5 cm and 4 cm shows a statistically difference in

Table 1. Mean (SD) peak joint reaction forces at knee and ankle segments
with respect to magnitude of the level of increment between the limbs

Level of LLD [cm]	JCF of Knee			JCF of Ankle		
	Short	Long	*P*-value	Short	Long	*P*-value
0.0	9.689456 ± 0.408038	9.757372 ± 0.442885	0.327	9.92007 ± 0.395372	9.989222 ± 0.448973	0.467
0.5	9.648375 ± 0.260372	9.785640 ± 0.175416	0.248	9.792225 ± 0.330849	9.8241 ± 0.321812	0.548
1.0	9.691 ± 0.568955	9.950725 ± 0.564962	0.513	10.043811 ± 0.359408	9.985844 ± 0.312259	0.895
1.5	9.842898 ± 0.4459143	9.695767 ± 0.4493706	0.317	9.638866 ± 0.800539	9.8273 ± 1.032872	0.513
2.0	9.780175 ± 0.409246	9.743335 ± 0.298272	1.000	10.01095 ± 0.405711	9.877016 ± 0.629398	0.749
2.5	9.801125 ± 0.278157	9.914459 ± 0.714191	0.564	10.13025 ± 0.5229192	9.998794 ± 0.46573767	0.317
3.0	9.732167 ± 0.210611	10.023932 ± 0.165539	0.127	10.144083 ± 0.446709	9.967344 ± 0.41130344	0.317
3.5	9.94798 ± 0.185663*	10.274778 ± 0.287449*	0.055	10.0943 ± 0.2019696*	10.25625 ± 0.39539*	0.200
4.0	9.733033 ± 0.143933*	10.227081 ± 0.284172*	0.010**	10.452666 ± 0.317885*	9.91575 ± 0.271429*	0.010**

Table 2. Mean (SD) peak joint reaction forces at knee and ankle segments
with respect to magnitude of the level of increment between the limbs

Level of LLD [cm]	JCF of Pelvis			JCF of Hip		
	Short	Long	*P*-value	Short	Long	*P*-value
0.0	9.206906 ± 0.413513	9.263333 ± 0.449106	0.429	9.170889 ± 0.417189	9.263333 ± 0.358964	0.393
0.5	9.19155 ± 0.252696	8.798525 ± 0.548161	0.386	9.145425 ± 0.262096	9.0101 ± 0.346191	0.564
1.0	9.3429 ± 0.435171	9.422433 ± 0.578712	0.827	9.314533 ± 0.341457	9.266 ± 0.3865714	0.627
1.5	9.341833 ± 0.4580622	9.209498 ± 0.4455811	0.317	8.819066 ± 0.830961	9.3164 ± 0.6732246	0.275
2.0	9.26315 ± 0.420129	9.23175 ± 0.328806	1.000	9.276366 ± 0.347372	9.331533 ± 0.33054	0.268
2.5	9.29035 ± 0.310694	9.3601 ± 0.715418	0.564	9.380911 ± 0.4669891	9.235022 ± 0.4778664	0.317
3.0	9.271167 ± 0.190783	9.520567 ± 0.141269	0.127	9.443806 ± 0.3540629	9.252011 ± 0.4114604	0.317
3.5	9.389217 ± 0.178181*	9.734983 ± 0.328713*	0.051**	9.349267 ± 0.179319*	9.695816 ± 0.320540*	0.053**
4.0	9.23305 ± 0.163279*	9.640366 ± 0.346861*	0.016**	9.181033 ± 0.1489342*	9.60415 ± 0.338517*	0.037**

Note: Results from Mann–Whitney *U*-test (within each level increments and long leg (LL) and short leg (SL)) are indicated by the symbols below:
* Significant ($P < 0.05$) there is statistical significant between level,
** Significant ($P < 0.05$) there is significant between both left and right.

value ($P < 0.05$). None of other level exhibits the same result ($P > 0.05$).

Subsequently, the ASI percentages of pelvic at 0 cm prior to 2 cm show no asymmetric difference between limbs (0–1%). However, the asymmetry between the limbs begin to ascend beginning on a level of 2.5 cm to 4 cm of discrepancy (2–6%). Thereafter, the statistical analysis is performed and the results are shown in Table 2. It can be noted that the only levels of 3.5 cm and 4 cm exhibits a significant difference between limbs ($P < 0.05$). None of other level shows the significant differences between both the limbs ($P > 0.05$). The ASI value for each of this level displayed an increasing trend. The most striking result to emerge from the data is that both pelvis and hip segment showed a higher mean value in tandem with the increment level of LLD, especially 4 cm level increase. These results are matched with the statistical method applied to identify the affected level, comparing 0 cm along the discrepancy level. It is presented that only 3.5 cm and 4 cm is statistically significant ($P < 0.05$). Others shows no statistical difference on level class.

Table 3. Symmetry indices for joint contact forces

Level of LLD [cm]	Symmetry Index Measure [%]			
	Pelvis	Hip	Knee	Ankle
0.0	0.07	0.08	0.70	0.69
0.5	0.93	0.39	1.41	0.32
1.0	0.85	0.41	2.64	0.28
1.5	1.43	1.85	1.51	1.37
2.0	1.80	1.81	1.65	1.16
2.5	2.86	1.94	1.15	1.29
3.0	4.03	4.00	2.95	2.18
3.5	4.64	5.80	3.23	1.59
4.0	6.61	6.40	4.95	2.53

4. Discussion

This study investigated the effect of different increment LLD's level on the lower extremity joints contact force during normal gait. There are no significant differences between both the short and long leg when the level is from 0.0 cm until 1.0 cm for most of joints. The rise in trend value could be seen starting from levels 1.5 cm until 4.0 cm. The results which demonstrated that the shorter leg induced more load is consistent with the study conducted by White et al. [9]. Their results showed that the increase in the contact forces is due to the increment in the weight acceptance forces as the transition of body weight from a greater vertical height (LLD level), also, the weight acceptance force rate increases as the existing adjustment in made to the joint kinematics [17], [21]. However, our finding also argued well with a recent study which said that an activity including a single leg stance exhibited greater forces [22], [23]. Maximum forces were recorded for the transferring motion for which main loading of a single leg was activated. Our data shows that subjects balanced the load status where by the ipsilateral leg carried nearly the whole weight, with the contralateral leg nearly unloaded.

The biomechanical structure of an ankle consists of a large weight-bearing area, compared to the other joints. This structure enables the ankle to spread load over a large area and to reduce stress across the joints.

Thus, the contact load on the ankle should be lower in value, compared to other joints. Whilst in our finding the ankle shows a large rise in value of contact forces magnitude, this is contradicting the explanation by Monk et al. [23]. In our study, we were comparing the ankle magnitude contact forces with an asymmetric index and the values does not indicate any symmetry change along the lower levels but begins to increase at the 4 cm of insole attachment. It was quite perceptible that even though ankles exhibit a large magnitude of contact force, it does not give any significant difference at the lower level of below 4 cm. This observation was then proven correct after statistical analysis was performed, there were also no significant differences at lower level and began to be after 4 cm level increase in insole height.

However, the knee joint shows a consistent increment difference between the two limbs, short limbs present with a higher magnitude result when compared to the long limb. The result obtained for the common affected limbs suggest that there is an increase risk factor for knee degerative joint disease on the shorter limb. These results mirror those of previous study, which examined the prevalent effect of LLD on knee osteoarthritis [24]. The focalization of increased risk factor in the short limb due to the biomechanical mechanism utilised by an individual to adapt to LLD conditions. These occurrences may be because the shorter limb has to travel distally from the ground, engage a higher impact on that side as a result of greater velocity and the likelihood of occurring “downhill” effect with each stride. Souza and Power also revealed that LLD induced inter-limb difference in hip muscle length, that follows may be a result of length-tension muscle properties that may change between two limbs at a specific hip angle [24]. This may result in altering mechanism of knee forces induced due to weakness at the functional hip. In addition, previous studies demonstrated that those with LLD level of greater than 1 cm are likely 1.5 times to develop knee arthritis due to the increase of pelvic obliquity and adduction moment on knee of affected limb [25].

Among the four joints, it is shown that the hip and pelvis joint were affected the most. It is based on the ASI percentages, which gave a large index percentage on hip and pelvic segments. The results matched with those in the previous study by Morscher, who described that the greater contact forces were demonstrated due to the surface area of the acetabulum cup of the hip feature [26]. This statement is in agreement among previous findings [10], [27], [28], which state that the joint contact forces magnitude was independent along the increment level of LLD. This result arises from the tilting structure occurring on the pelvic segment, then making the pelvic obliquity result in a posterior rotation, and subsequently shifting the centre of the body mass towards the shorter side.

We evaluated that the model used in this study was adequate for the current research question. Yet, there are some limitation that should be considered. We used only one model to scale each of the participant subjects. The scaling model was based on anthropometric data of each subject (height and weight). This only one model was used to avoid any error being made by creating different model as the number of subjects were large. Apart from that, we cannot provide any statistical evidence in this study. Thus, the result presented is limited to discussion on mean value without any of it statistically proven. In addition, we limited the gender of participants by using only male subjects. This is because we want to minimize the variables arising during data analysing. There is no evidence of to what extend the gender distribution would affect the finding, hence, the comparison between genders will be further analysed in the future.

5. Conclusions

Contact forces for four joints were recorded for eighteen normal male subjects. All joints contact forces showed no significant changes, based on the ASI values, which is zero or nearly zero with the elevation level 0.0 cm until 1.0 cm, while there are significant differences noted as the level increases to levels from 1.5 cm to 4.0 cm. The forces distributed on the long legs seem relatively independent of the LLD height, due to fluctuation in force magnitude. Shorter legs carry more force as compared to the long leg. The most positive increment between each level can be seen on the hip and pelvis segments. The primary objective of this study was to gather the information on which joint and limb were most affected during walking, as there are arising levels of LLD. The result obtained hereby might help any clinician that treats a patient with a structural LLD. When a greater force was concentrated on the specific joint, the joint would later develop a joint disease, such as osteoarthritis.

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