Gait & Posture 67 (2019) 71–76

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Dynamic evaluation of simulated leg length inequalities and their effects on the musculoskeletal apparatus

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ARTICLE INFO

Keywords:
Rastereography
Leg length discrepancy
Leg length inequality
Static
Dynamic
Gait
Surface topography

ABSTRACT

Background: Leg length inequalities (LLI) are a common problem in medicine. So far, the diagnosis and treatment are performed under static conditions. Surface Topography (ST) is an optical, non-invasive technique that uses the principle of triangulation to measure spinal posture and pelvic position. This technique offers the opportunity to detect and treat LLIs and their effects under dynamic conditions.
Research: question The aim of the study is to show that ST can detect simulated LLIs under dynamic conditions and to prove if there are differences between the effects on the human body under static and dynamic conditions.
Methods: In the clinical study a total of 30 test subjects were examined with a ST measuring device. LLIs (1 to 4 cm) were simulated using a custom- built sandal and insoles of various thickness. The pelvic obliquity, the surface rotation and lateral deviation of the spine were detected on a treadmill under static and dynamic conditions (3 km/h).
Results: Under static and dynamic conditions LLIs lead to a significant increase of all measured parameters. The pelvic obliquity reaches a significant level of $p < 0.0001$ under static and $p = 0.0001–0.0421$ under dynamic conditions. However, for all examined parameters the magnitudes of the parameters under dynamic conditions were smaller than under static conditions.
Significance: The study showed that simulated LLIs also have a significant effect on the human pelvis and spine under dynamic conditions, but with a smaller magnitude than under static conditions. The human individuum is a dynamic one. Because of that, for the future it should be of great interest to use dynamic measurements to detect and treat LLIs to provide an over correction of LLIs.

1. Introduction

Leg length inequality (LLI) is a condition in which paired limbs are noticeably unequal [1]. They can be found in 40–70% of the population and they may be greater than 2 cm in about 0.1% [1]. LLIs can be a predisposing factor for sacroiliac joint disorders, lumbar back pain, functional scoliosis as well as for symptoms in various joints due to the kinematic joint chain [2]. LLIs can affect all age groups and are categorized into anatomical and functional LLIs [1].

So far, the clinical diagnosis and treatment of LLIs is still performed mostly under static conditions, while patients are standing upright in front of the examiner. The amount of LLI is measured by palpating the height of the iliac crests and their position to each other. Larger LLIs lead to pelvic obliquity, which is then corrected by placing small blocks under the short leg until the pelvis is levelled. In multiple studies, the static effects of LLIs on the musculoskeletal system have been evaluated [3–5]. Hackenberg et al. [5] showed that there is a direct effect of simulated LLIs on the pelvic position and spinal posture [5]. Betsch et al. established a non-invasive method to simulate and examine LLIs and their effects on the musculoskeletal apparatus using a simulation platform [3,4]. The results of these studies confirmed a correlation between increasing LLIs and pelvic obliquity, torsion and changes of the spinal posture [3,4].

The human being is a dynamic individual [6] and therefore, we believe that the diagnosis and treatment of LLIs should also be carried out under dynamic conditions. With the development of surface topography there is a fast, reliable and radiation-free method available to diagnose and treat LLIs under dynamic conditions. In previous studies, the reliability and validity of this system under static and dynamic conditions was shown [7–10].

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https://doi.org/10.1016/j.gaitpost.2018.09.022
Received 16 April 2018; Received in revised form 18 August 2018; Accepted 23 September 2018

Fig. 1. We constructed a custom sandal in three different sizes. To simulate the different LLIs, three insoles with varying thickness, were added to achieve a maximum LLI of 4 cm.

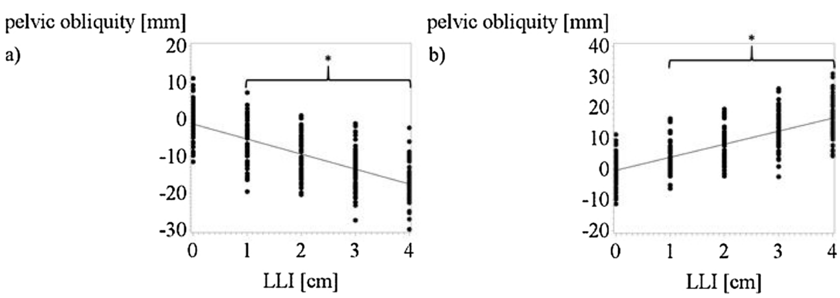

Fig. 2. Pelvic obliquity in mm. a) shows the effect of a simulated LLI on the left side. b) shows the effect of a simulated LLI on the right side. Negative values indicate a pelvic obliquity to the left side and positive to the right side. The brackets mark the LLIs, which reach the level of significance [$p < 0.05$] in relation to the reference LLI (0 mm). The pelvic obliquity was directed towards the longer leg side. The reported dots shown in this figure represent one single trial.

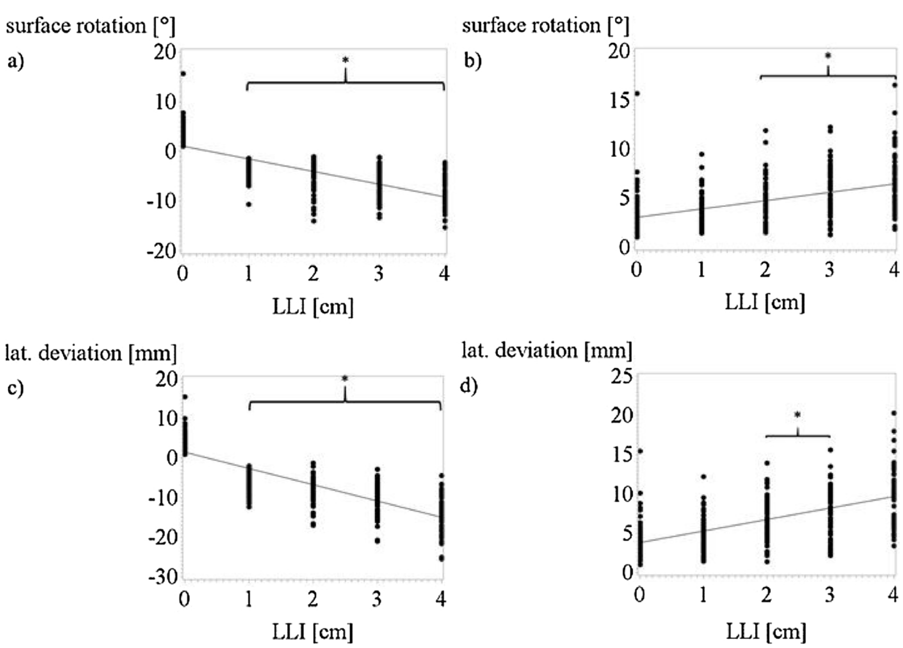

Fig. 3. Surface rotation in ° and lateral deviation in mm. a) and c) shows the effect of a simulated LLI on the left side. b) and d) shows the effect of a simulated LLI on the right side. Negative values indicate a surface rotation to the left side and positive to the right side. The brackets mark the LLIs, which reach the level of significance [$p < 0.05$] in relation to the reference LLI (0 mm). The surface of the back is rotated and laterally deviated to the side of simulated LLI. The reported dots shown in this figure represent one single trial.

Aim of this study was to develop a method for simulating and evaluating LLIs and their effects on the pelvis and spine under dynamic conditions.

2. Materials and methods

30 test subjects without pre-existing leg or spinal abnormalities were included in this study. Exclusion criteria of this first pilot study were a pelvic obliquity due to a functional or anatomical leg length discrepancy greater 10 mm and obesity with a body mass index (BMI) of greater 35 kg/m^2, which could impede the detection of anatomical landmarks by the measuring system. Another exclusion criterion was back pain during the previous year lasting longer than 2 days. Mean age of the subjects was 24.4 years (± 2.2 years), mean height was 1.77 m (± 0.08 m) and mean weight was 70.5 kg (± 7.2 kg). The subjects gave their oral and written consent to participate in this study. The protocoll of this study was approved by the local ethic commity (Study number: 111-15).

All measurements were conducted with a surface topography measuring system (Formetric 4D motion, Diers International GmbH, Germany). This system allows a radiation- and contact-free examination of the pelvic position and spinal posture [11]. It consists of a light projector and a digital video camera, which work together as a stereooptical pair [8]. The method of surface topography is based on the mathematical principle of triangulation, which uses the projection centers of the stereooptical pair to measure the 3D coordinates on an object's surface [12]. The projector generates a raster image on the back, which is captured by a digital camera [12]. The computer unit

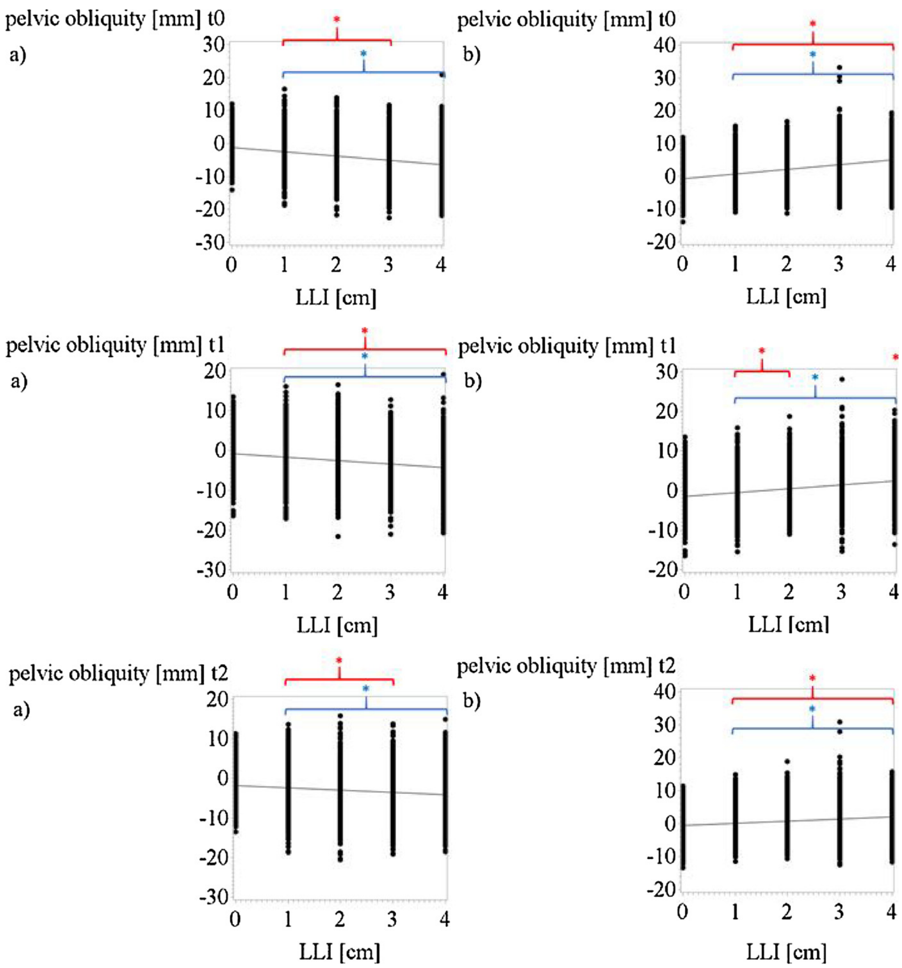

Fig. 4. Pelvic obliquity for each gait phase t0, t1 and t2 in mm a) shows the effect of a simulated LLI on the left side. b) shows the effect of a simulated LLI on the right side. Negative values indicate a surface rotation to the left side and positive to the right side. The brackets mark the LLIs, which reach the level of significance [$p < 0.05$] in relation to the reference LLI (0 mm) (blue: left foot, red: right foot). The pelvic obliquity was directed towards the longer leg side and showed increasing values with increasing of the LLI. The reported dots shown in this figure represent one single trial. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

then creates a three-dimensional model of the human backshape and spine by analysing the convex, concave and saddle-shaped areas of the back surface [13]. These areas are linked to specific anatomical landmarks of the human spine and pelvis [12]. Based on the surface information of these specific anatomical landmarks and in combination with biomechanical modelling, the geometry of the vertebral bodies and pelvic bone is calculated. Consecutively, the shape of the spinal column is animated in 3D with its frontal, sagittal and transversal plane characteristics, based on a model created by Turner-Smith [14,15]. It is also possible to use the two lumbar dimples to measure pelvic obliquity based on their close relation to the underlying posterior superior iliac spines [12,16]. In a series of studies, rasterstereography has proven to have high reliability and accuracy, when compared to X-rays [9,17–19].

All measurements were performed while the subjects were standing or walking on a treadmill. The respective leg length differences were simulated by a custom- built sandal with insoles of various thickness (1–4 cm) (Fig. 1). We chose to use a sandale for simulating LLIs instead of regular shoe insoles, because we wanted to accommodate for any weight differences and differences in foot wear between the subjects.

The static measurements were done in an upright position with extended knees and arms hanging on the sides, as previously described (Fig. 2) [4]. All test subjects stood for sixty seconds to adapt to the simulated LLIs prior to the measurement. The weight distribution between both legs during static trials was measured by a pressure plate built into the treadmill to ensure an equal weigth distribution (± 5% side differences). First, base line measurements without the sandale were performed. Then, LLIs on the right then on the left side were simulated by adding insoles from 1 to 4 cm into the sandal.

All dynamic measurements were performed while the subjects were walking on a treadmill with a velocity of 3 km/h. To compensate for differences in the number of gait cycles recorded caused by height differences of the test subjects, we used the first four complete gait cycles for analysis in every subject. Follwing the same protocol as during the static measurements, we first measured the spinal posture and pelvic position without wearing the sandal followed by insoles from 1 to 4 cm on the right then on the left side.

For the purpose of this study it is necessary to define certain terms regarding the parameters that were measured. Pelvic obliquity is the amount of tilt in millimetres from the horizontal line between the two lumbar dimples DL (left dimple) to DR (right dimple). A positive value indicates that the right dimple is higher than the left and a negative value indicates that the left dimple is higher than the right. Surface rotation is defined as the vertebral rotation measured perpendicular to the back surface over the spinous processes as the central tendency from the spinous process of the 7th cervical vertebra to the midpoint between the two lumbar dimples measured in degrees (°). Lateral deviation is defined as the deviation of the spinal midline from the line between the spinous process of the 7th cervical vertebra (VP) to the midpoint between DL and DR (DM) in the frontal plane [4].

2.1. *Stastical analysis*

All data were checked for Gaussian distribution by the Chi- square test of normality and presented as means with standard deviations or 95% confidence levels. Student *T*- tests were performed to check for differences between the different LLIs. The level of significance was set at $p < 0.05$. For the statistical analysis of the dynamic data we analyzed three gait phases acoording to Perry et al. [20]. The first time

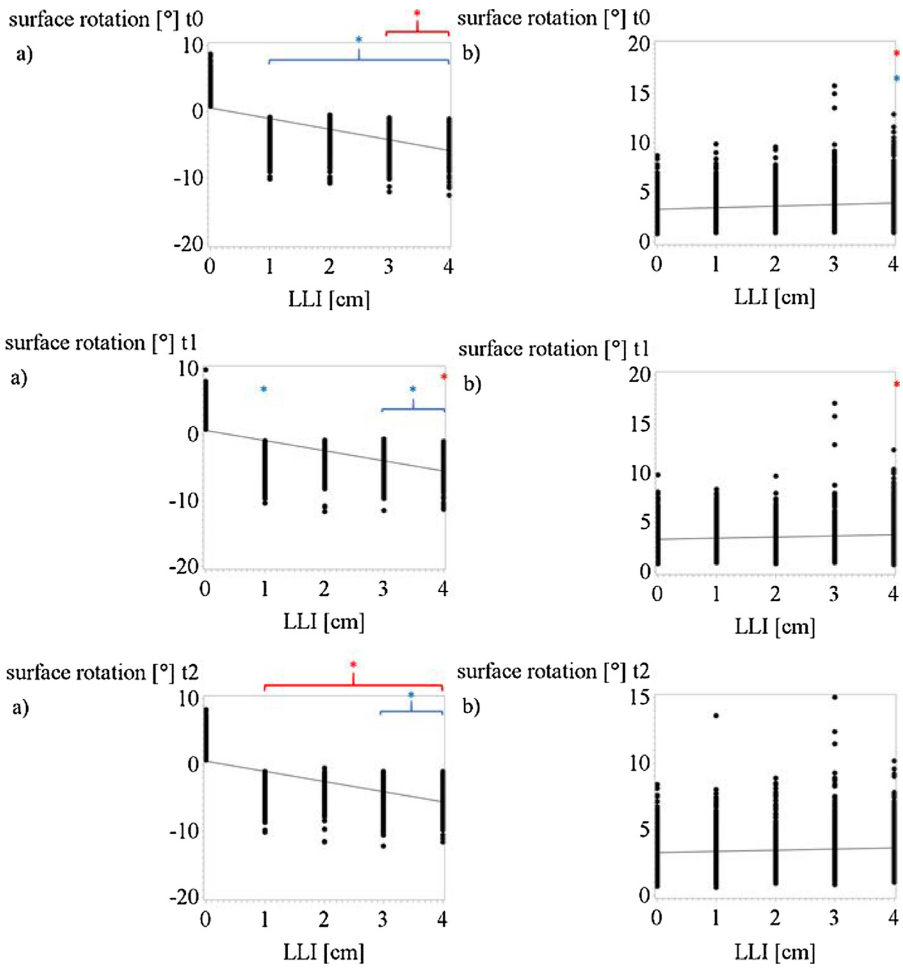

Fig. 5. Surface rotation for each gait phase t0, t1 and t2 in ° a) shows the effect of a simulated LLI on the left side. b) shows the effect of a simulated LLI on the right side. Negative values indicate a surface rotation to the left side and positive to the right side. The brackets mark the LLIs, which reach the level of significance [$p < 0.05$] in relation to the reference LLI (0 mm) (blue: left foot, red: right foot). The surface of the back was rotated to longer leg side where the LLI was simulated. The reported dots shown in this figure represent one single trial. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

point of the stance phase was the initial contact of the foot (t0), the second was the midstance phase (t1) and the final phase was the toe off (t2). Statistical analysis and graphics were prepared with the SAS (SAS 9.4 SAS Institute Software GmbH, Wien, Austria) and Microsoft Excel (Microsoft Excel 2016, Microsoft Corporation, Redmond, USA).

3. Results

3.1. Static measurements (all results are presented in relation to the reference condition LLI = 0 mm)

Under static conditions a simulated LLI of 1 cm and greater led to a significant increase of the pelvic obliquity ($p < 0.001$) (Fig. 2).

The results also showed that with simulated LLIs of 1 cm and greater on the left and 2 cm and greater on the right side significant changes of the surface rotation occurred (left sided LLI: p = 0.0001–0.0229, right sided LLI: p = 0.0002–0.00016) (Fig. 3).

We also did find a significant increase under static conditions for the lateral deviation of the spine with simulated LLIs of 1 cm and greater ($p < 0.0001$) on the left and 2 cm and greater (p = 0.0001–0.041) on the right side (Fig. 3).

3.2. Dynamic measurements (all results are presented in relation to the reference condition LLI = 0 mm)

In contrast to previous studies, we have also simulated LLIs by using a custom-built sandal with different insoles, while the test subjects were walking on a treadmill. We found a significant increase of pelvic obliquity with increasing LLIs (Fig. 4). The hemi-pelvis of the longer leg is higher than the contralateral hemi-pelvis of the shorter leg. These changes were significant for left sided LLIs of 1 cm and greater during the stance phase of the left foot (t0: p = 0.0001–0.0002, t1: p = 0.0001, t2: p = 0.0003–0.0178) and during the stance phase of the right foot of 1 cm–3 cm in t0 (p = 0,0176–0,0481) and t2 (p < 0,0001–0,0001) and for 1 cm in t1 (p = 0,0107). Right sided LLIs of 1 cm and greater (t0: p < 0,0001–0,001, t1: p < 0,0001, t2: p < 0,0001–0,0073) induced a significant pelvic obliquity during the stance phase of the left foot. While standing on the right foot right sided LLIs led to significant changes for 1 cm and greater in t0 (p = 0.0025 – 0.0494) and t2 (p < 0.0001 – 0.001) and for 1 cm, 2 cm and 4 cm in t1 (p = 0.0002 – 0.034). (Fig. 4).

The dynamic experiments showed also an increasing surface rotation with higher LLIs during all captured time points (Fig. 5). The surface rotation of the spine is orientated towards the longer leg. These effects showed significant differences compared to the neutral standing position for the stance phase of the left foot for a left sided LLI of 1 cm for t0 ($p < 0.0001$) for 1 cm, 3 cm and 4 cm for t1 (p = 0.0014 – 0.0187) and for a LLI of 3 cm for t2 (p = 0.0002 – 0.0122). During the stance phase of the right foot LLIs of 3 cm in t0 (p = 0.0001 – 0.0392), of 4 cm in t1 (p = 0.0162) and of 1 cm in t2 (p = 0.0001 – 0.0004) reached the level of significance. A right sided LLI of 4cm showed significant effects for the gait phase t0 of the left and right foot (left foot: p = 0.0335, right foot: p = 0.001). In the phase t1 there were only significant results for a LLI of 4 cm in the stance phase of the right foot (p = 0.0158) (Fig. 5).

The lateral deviation showed similar changes. We found an increasing lateral deviation towards the longer leg with increasing LLIs (Fig. 6). The level of significance was reached for left sided LLI and the

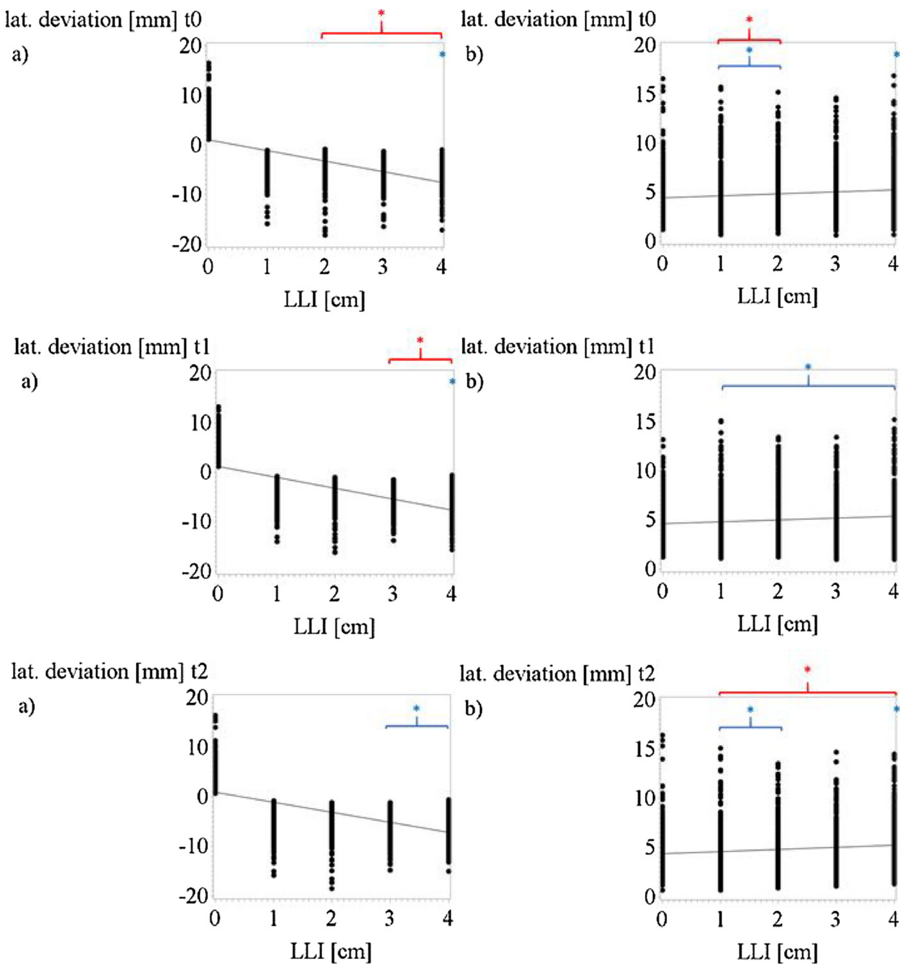

Fig. 6. Lateral deviation for each gait phase t0, t1 and t2 in mm a) shows the effect of a simulated LLI on the left side. b) shows the effect of a simulated LLI on the right side. Negative values indicate a lateral deviation to the left side and positive to the right side. The brackets mark the LLIs, which reach the level of significance [$p < 0.05$] in relation to the reference LLI (0 mm) (blue: left foot, red: right foot). The lateral deviation of the spine increased on the longer leg side where the LLI was simulated. The reported dots shown in this figure represent one single trial. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

stance phases of the left foot of 4cm for t0 ($p = 0.0214$) and t1 ($p = 0.0197$) and of 3 cm for t2 ($p < 0.0001$). The stance phases of the right foot led to significant differences in t0 for a LLI of 2 cm ($p < 0.0001 – 0.0291$) and in t1 for a LLI of 3 cm ($p < 0.0001 – 0.0004$). Right sided LLIs showed significant effects in the stance phase of the left foot in t0 and t2 for a LLI of 1 cm, 2 cm and 4 cm (t0: $p = 0.0005 – 0.0385$, t2: $p = 0.0041 – 0.0134$). Furthermore, there were significant changes in t1 for a LLI of 1cm ($p < 0.0001 – 0.0043$). The stance phases of the right foot showed only significant effects (t0: $p = 0.0003 – 0.0004$, t2: $p < 0.0001 – 0.0358$) for 1 cm and 2 cm during the initial contact and for a LLI of 1 cm during the toe off (Fig. 6).

4. Discussion

Main goal of this study was to develop a method that allows the evaluation of LLIs under dynamic conditions. In order to be able to simulate and correct LLIs, we built three sets of sandals with insoles of varoius thickness (1–4 cm) that can accommodate most feet. We chose to use this model, instead of using regular shoe insoles with varying thickness, because we wanted to be able to simulate LLIs independent of the subjects anatomy, body weight and type of shoe wear.

The results of this study indicate that simulated LLIs have similar effects during walking than previously published during static standing [4]. In general, we found that the magnitude of the effects on pelvic position and spinal posture were smaller during walking on a treadmill than while standing. We believe that some of the differences can be explained by the standing position during the static measurements. During the static experiments, all subjects were instructed to stand upright with a relaxed posture and fully extended knees (Fig. 2). This position does not allow for any compensation of the simulated LLIs at the ankle and knee joints. In contrast, during the dynamic measurements on the treadmill all subjects were allowed to walk freely, which does also allow a compensation at the ankle and knee joints. This could explain the smaller magnitude of changes found during the dynamic measurements at the pelvic and spinal level. Walsh et al. [21] confirmed these findings in their study examining seven subjects while simulating LLIs of up to 5 cm [21].

In our study, we could see differences between the effects of simulated LLIs on the right and left foot during gait. For the parameter surface rotation, simulated LLIs on the left side induced slightly greater effects on the muskuloskeletal system then on the right side. This could be the effect of the dominant leg. There are studies indicating a side difference between the dominant and non-dominant leg [22]. However, these side differences were only found for one parameter. Nevertheless, in future studies, it would be of clincal interest to test and evaluate for a dominant leg effect.

An advantage of this study is the use of surface topography to measure the effects of simulated LLIs on the pelvic position and spinal posture. This technique has shown its high validity and reliability in numerous studies under static as well as under dynamic conditions [7–10]. The use of surface topography does allow to measure leg length inequality indirectly, by quantifying the amount of pelvic obliquity. With the use of a simulation platform it is possible to compensate for leg length differences and then evaluate their effects in real-time on the spinal posture and pelvic position. The chosen leg length compensation can then be directly tested under dynamic conditions using our custom-built sandal with the appropriate insole thickness.

We believe that the results of this study could be of great clinical importance, since humans are dynamic individuals, so LLIs should also be treated under dynamic conditions. In various studies, it was shown that LLIs can have a significant impact on the musculoskeletal apparatus. For example, LLIs can lead to hip pain in the longer leg [1,23,24]. This can be explained by an increase in pressure between the femoral head and the acetabulum caused by pelvic obliquity [23,25,26]. A further effect of LLIs is a decrease in the center edge angle of the hip joint, which can increase the risk for a femoro-acetabular impingement [25,27]. LLIs can also cause low back pain, which has been shown in multiple studies [2,23,24]. LLIs negatively affect the spine by causing a lumbar scoliosis, which is directed towards the shorter leg [1,2,23,28,29].

A limitation of this study is that we have simulated acute LLIs instead of evaluating long- standing “real” LLIs. We believe, it is important to first test a new method under laboratory conditions before evaluating actual patients. Furthermore, we chose to use three custom-built sandals to simulate LLIs, which covered most but not all shoe sizes. That may have led to problems with the footwear, in particular when LLIs of 3 and 4 cm were simulated. The use of surface topography made it necessary to conduct all measurements while the subjects were walking on a treadmill. In previous studies, it was show that there do exist differences between walking on a treadmill and walking on normal grounds. Riley et al. [30] were able to show that healthy subjects walked on a treadmill with slower maximum velocity as compared to walking on normal grounds. We believe that in follow-up studies it would be of interest to measure patients with “real” LLI by using surface topography.

5. Conclusion

LLIs can lead to multiple musculoskeletal disorders. The treatment and diagnosis of LLIs is still carried out under static conditions, although patients are moving and walking most of the time. We were able to develop and evaluate a novel method to simulate and examine the effects of LLIs on the musculoskeletal apparatus under dynamic conditions.

Conflict of interest

None

Declaration of interest

None.

Acknowledgments

All procedures performed in studies involving human participants were in accordance with the ethical standards of institutional and/or national research committee and with the 1964 Helsinki declaration and its later amendments or comparable ethical standards. Informed consent was obtained from all individual participants included in the study. This research did not receive any specific grant from funding agencies in the public, commercial, or not-for profit sectors.

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