mobiLLD: Exploring the Detection of Leg Length Discrepancy and Altering Gait with Mobile Smart Insoles

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Figure 1: A vast group of people, ~23% of the world population, suffers from significant Leg Length Discrepancy (LLD) of 10mm or more [35]. This condition causes uneven gait, often resulting in health issues, including spinal pain and headache. We explore the detection of LLD using a mobile smart insole system. We use Inertial Measurement Units (IMU) and pressure-sensitive insoles to measure gait parameters, such as Stance Time, Ground Reaction Force (GRF), and Center of Pressure (CoP). To alter asymmetrical gait, we augment vibrotactile feedback under the foot.

ABSTRACT

Leg length discrepancy (LLD) is common and typically burdens the spine and hip causing a variety of health issues including back pain and headache. Common orthopaedic solutions target correcting gait, typically by shoe lifts. Moreover, functional LLD can be temporary as it occurs after an injury or even after sitting in a twisted posture. Often, it is only noticeable when pain has arisen. We present a mobile smart insole system designed to detect LLD by measuring gait parameters, such as Stance Time, Ground Reaction Force, and Center of Pressure. Furthermore, our prototype is capable of augmenting vibrotactile feedback under the foot. Our method has shown to impact gait, in particular Stance Time, which may be used to compensate gait asymmetries caused by LLD. To evidence our findings, we rely on a similar methodology from related lab studies and induced a mild LLD by a 10mm offset insole among 16 participants.

CCS CONCEPTS

• Human-centered computing → Ubiquitous and mobile computing; • Applied computing → Health care information systems.

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1 INTRODUCTION

Leg length discrepancy (LLD), also known as anisomelia, is defined as a condition in which paired lower extremity limbs have noticeably unequal length [21]. LLD is a common abnormality affecting 65% and 90% of the population at varying severities [28]. In most cases, the body unconsciously compensates for this issue enabling us to walk straight. However, ~23% of the world population suffers from significant LLD of 10mm or more [35], causing an uneven gait. Overcompensation for an uneven gait can result in injuries, balance issues affecting stride, and other long-term effects [17, 18, 21]. Studies have also shown that LLD is a predisposing factor for associated musculoskeletal disorders [18, 40]. Further, LLD has been implicated in affecting standing posture, increasing the incidence of scoliosis, lower back, hip, and spine pain, and lower extremity stress fractures [29]. The existence of LLD impedes humans, thus creating the need to detect and treat the condition.

In contrast to severe LLD, which is often anatomical or structural LLD (SLLD), research has shown that LLD can be functional (FLLD) [9]. A joint contracture, static and dynamic mechanical axis malalignment, muscle weakness, or shortening are possible causes of FLLD [9, 13, 27]. FLLD can even occur temporarily, namely when posture is poor while sitting [21]. The immediate consequence is often spinal pain and headache. Therefore, detecting this type of LLD early and deploying a compensation mechanism is desirable.

Our research addresses such issues by deploying a mobile detection method based on pressure-sensitive insoles and IMUs that can be unobtrusively embedded into our everyday life. We extract three gait parameters: Stance Time, Ground Reaction Force (GRF), and Centre of Pressure (CoP). Abnormalities in gait, such as caused by LLD, can be observed best when removing a person's visual perception. When blindfolding we expect a walking person to curve due to asymmetrical gait [46]. Such walking deviation can be used as another indicator for LLD. In our research, we aim to explore a mobile detection approach to determine LLD. Finally, we augmented feedback under a single foot by providing a vibrotactile stimulus, hypothesising to alter gait, and ultimately to compensate for LLD. To validate our hypothesis, we ran a user study with 16 participants. Since there is no prior work on our newly proposed method, the ethics committee only approved justifiable means in the form of controlled lab studies with low risk to "Construct Validity" [10]. In non-clinical studies, we are not permitted to include participants with medical conditions to prove "Ecological Validity" [10]. Therefore, we artificially induced an LLD by inserting a flexible 3d-printed insole at a single side, resulting in a 10mm discrepancy. This is a common method seen in many other clinical studies [5, 15, 33]. Our lab study is a necessary first step to explore and uncover the potential and technical feasibility of such a new approach to detect LLD and alter gait for possible intervention.

Our study could confirm previous knowledge, showing that an induced artificial LLD can negatively impact gait, visual perception can compensate for discrepancies, and blindfolding can disarm our visual compensation mechanism. Furthermore, our study contributes the following unique findings:

- pressure-sensitive insoles with IMUs can be used as an unobtrusive mobile sensing method to detect gait parameters and possible asymmetries that LLD may cause,
- augmenting vibrotactile feedback under the foot can alter gait, which may be used to compensate LLD.

2 RELATED WORK

2.1 Detection and Measurement of LLD

2.1.1 Non-Clinical Methods. Radiography is considered the gold standard for measuring LLD; there are three methods that utilise this technology to measure LLD, all measuring from some landmark on the proximal femur to somewhere on the ankle, ignoring foot to limb length [11]. These other methods include 'orthoroentgenogram' [49], 'scanogram' [21] and 'computerised digital radiograph' [37]. Other methods of measuring LLD include computerised tomography (CT) [1], three-dimensional ultrasonography (3-D US) [26], and magnetic resonance imaging (MRI) [21].

2.1.2 Clinical Methods. There are two main clinical methods used to measuring LLD, an *'indirect method'* and a *'direct method'*. The indirect method is done standing using lift blocks under the shorter leg and visually examining the level pelvis [53]. The indirect method is done in the supine position [53], measuring the distance of fixed bony landmarks with measuring tape [21]. Other researchers concluded that the validity is as satisfactory and reliable when used as a screening tool [6, 20, 22].

2.1.3 Alternative Methods. According to Schaeffer [46], humans tend to turn in one direction when walking. An explanation for this tendency is due to biomechanical asymmetries, such as LLD [30]. A study was done to test for general directional bias where blindfolded participants were told to walk straight in the direction indicated to them at the beginning [48]. For short distances, it was found that blindfolded people showed a small amount of veering, but for large distances, veering could not be quantified due to the accumulation of sensory noise [48]. Other studies concluded that although biomechanical asymmetries could influence veering, it is more likely that curved walking is related to postural performance [7]. Thus, the magnitude of veering may indicate that LLD is present, but other factors could also impact.

Research has yet to demonstrate how any type of detection method can be mobile and unobtrusively embedded into our everyday life, particularly to detect temporary FLLD, which can occur after sitting in an unfavourable posture for a prolonged period. We believe this lack of research exists because mobile technology has not yet considered an effective treatment of LLD.

2.2 Treatment of LLD

There are two common ways known to treat LLD, surgical and non-surgical. Mild cases of LLD are usually left untreated or treated non-surgically [37]. Treatment of moderate cases of LLD is dealt with on a case by case basis while severe cases should be corrected using surgical methods [45].

2.2.1 Surgical Treatment. Epiphysiodesis is a surgical method of treating LLD [14]. Physeal stapling is another method where staples are placed across the tibia on the medial growth plate to halt bone growth [12]. These methods reported good results in children in managing knee deformities [36]. However many complications with these procedures include overcorrection, as well as rebound longitudinal growth after removing the staples [21, 37].

2.2.2 Non-Surgical Treatment. The most common non-surgical treatment of LLD are: Prosthetic Fittings, Exoskeletons, Shoe Lifts, Orthopaedic Shoes and Insoles. We are having a closer look at Shoe Lifts, and Orthopaedic Shoes & Insoles as these are most common:

Shoe Lifts Shoe lifts are typically used as a treatment for discrepancies of less than 20 mm [44]. Internal heel lifts, internal shoe inserts, and external heel lifts are various shoe lift options. Internal shoe lifts are most frequently 5-15mm, while external heel lifts tend to be used for LLD of 15-20mm due to increased patient comfort [44]. Using heel lifts on patients with LLD of more than 10mm has been found to reduce lower back pain and increase the range of motion of the lumbar spine [17–19].

Orthopaedic Shoes & Insoles Custom-made orthopaedic shoes involve the modification of the shoe sole or the upper of the ready-made shoe and can benefit patients with a wide variety of pathologies through improving the patient's gait [47]. They are most commonly prescribed to patients with diabetes, rheumatoid disorders, and muscle disorders to help prevent foot ulcers, reduce pain, support anatomical foot deformities, and enhance stability [38]. Terrier et al. reported orthopaedic shoes to significantly decrease pain levels, by 29% on a visual analogue scale, for patients with complex foot and ankle fractures, and to improvee local dynamic stability by 7-10% in the medio-lateral, vertical, and anteroposterior axes [51]. An orthopaedic insole has similar benefits to orthopaedic shoes and can be prefabricated or individually made for the patient's foot [47]. A new approach to developing customised insoles, discussed by Peixoto et al., involves using a 3D scanner to collect data on the patient's foot to produce a 3D printed insole which reduces fatigue and is 25% lighter than conventional footwear [41].

The drawback of all types of assistive devices is the need to be individually manufactured and adjusted in a costly making process. Another problem underlying these devices is the individual wear-off. After a while, patients need to return to their orthopaedist to get further individual modifications of their assistive device. In fact, many orthopaedic solutions target changing the gait to compensate for LLD. Another method to alter gait is augmenting feedback, which has been widely explored with visual and audio aids over the past few decades [3]. Rather recent research has demonstrated the use of vibrotactile feedback at the arm [32], neck and shoulders [31], belly [52], and legs [2] to impact gait. Researchers performed these investigations with a variety of bulky and tethered prototypes to support stroke rehabilitation. Another tethered shoe prototype has been recently developed revealing how to improve body posture for standing exercises, while vibrotactile feedback stemming from the shoe wall is visualizing the Center-of-Pressure [16]. Yet, current research has not demonstrated how a truly mobile solution can look and how it can compensate for LLD.

3 MOBILLD

Current solutions to detect LLD usually require a physician. Accurate solutions are considerably costly. In contrast, inexpensive methods can be considered inaccurate. On top of this, no solution can actively balance or compensate LLD dynamically that tailors to the patients' daily conditions. As temporary FLLD can even occur after sitting in a twisted posture, or after an injury, etc., it often immediately results in spinal pain and headache. Therefore, having an assistive device capable of detecting and compensating LLD immediately would be useful.

To advance the state-of-the-art, we developed a fully mobile prototype – a pair of insoles *(see Figure 2)* that can be inserted into any ordinary shoe *(see Figure 3)*. The prototype for each foot consists of a pressure-sensitive insole (sensing.tex) and an IMU to detect LLD. Moreover, each insole incorporates 8 vibration motors embedded in a flexible 3d-printed insole (based on NinjaFlex material). The system is self-contained and driven by an Arduino Teensy 3.6, which is held within a 3D printed casing (together with a battery,

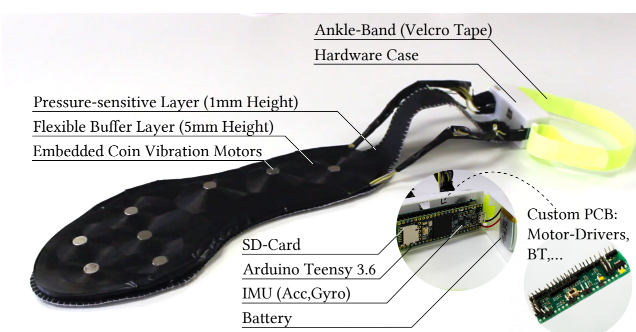

Figure 2: Depicting the insole prototype (right foot). The device is battery-powered and fully mobile. The insole is placed into a shoe, while the electronics is stored in casing that is attached to the ankle by Velcro tape.

Figure 3: mobiLLD prototype worn with sport shoes.

Bluetooth modem, SD-card, and a custom DC motor-driver board) that is strapped to the user's ankle with Velcro tape.

3.1 Implementation

3.1.1 Stance Time (ST). The stance phase refers to the stage of the gait cycle where the foot is in contact with the ground, as shown by Khamis et al. [27]. In contrast, the stride time is the time between two consecutive heel strikes by the same leg. The stance phase occurs from initial heel contact to the pre-swing stage where the toe lifts off the ground. We refer to the Stance Time for a leg as the duration in which the person is in the stance phase for a single gait cycle. Bhave et al. [8] found that people with gait asymmetries spend more time in the stance phase with their longer leg as the shorter leg shows a decreased Stance Time and a decreasing walking velocity. The resulting difference in Stance Time would also be pronounced with LLD, we believe.

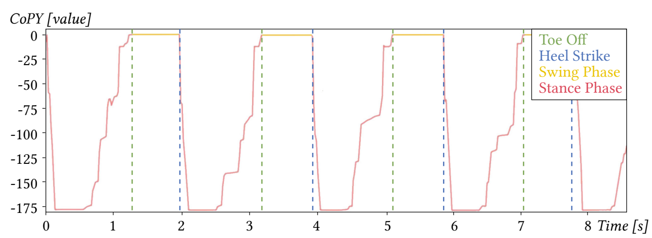

Figure 4: Gait cycle shown by CoP y-axis values. The Stance Phase begins with a Heel Strike and ends by lifting off the toes. Using the aforementioned calculation, the Swing Phase would not be calculated as division by zero is undefined.

Each pair of heel and toe contact points can be used to identify the start and stop times of each stance phase while a participant is walking. For N identified pairs, we can compute the Stance Time, ST_i for pair i as the difference in heel contact time, HC_i , and toe contact time, TC_i:

$$ST_i = TC_i - HC_i, \qquad for\ i = 1, \ldots, N$$

Using the data from both legs for a run, we can compute the difference in mean Stance Time between the right and left leg. We compute the stance time from the pressure insole and IMU data. Figure 4 depicts the gait cycle for the pressure insole. Another way to determine the Stance Time is using the IMU sensor, in particular the Accelerometer or Gyroscope. This method has been broadly used in research already. Figure 5 shows the gait cycle of a single axis from a Gyroscope attached to the ankle.

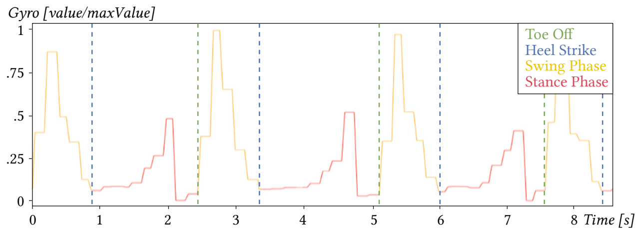

Figure 5: Gait cycle shown by angualar rotation values. The stance phase begins with a heel strike and ends by lifting off the toes. During the stance phase, rotation is marginal.

3.1.2 Ground Reaction Force (GRF). Refers to the force from the foot striking the ground. This force occurs at the initial heel contact, the first stage of the gait cycle as shown by Khamis el al. [27]. GRF is also known as, 'jerk,' which can be denoted as the rate of change of acceleration [24]. Studies show that differences in GRF between legs can occur, being a strong indication of gait asymmetry [21, 42]. Pereira and Sacco performed a study on LLD, evidencing the differences in GRF for both legs [29]. Research showed that GRF is larger on the longer leg [8, 23]. Therefore, comparing GRF of both feet is imperative when aiming to detect LLD.

Calculating GRF is also possible through using a pressure sensitive insole or an IMU. Since the hardware-driver of our sense.tex insole already somewhat normalizes the sensor readings, we decided to rely on an IMU. A previous study already found the IMU's Accelerometer to be reliable. We followed Kawamura et al. [25] who found that acceleration of a single axis - upper direction (z-axis) to be sufficient for calculating the GRF:

$$GRF_t = \frac{A_{z,t} - A_{z,t-1}}{\Delta t}$$

Once we found the mean GRF for each leg, we calculated the percentage difference in GRF between both legs.

3.1.3 Center of Pressure. There is evidence that as a compensatory mechanism to LLD, the longer leg adapts by pronation of the foot, leading to supination on the shorter leg [29]. We believe that this mechanism will be evident in differences in Center of Pressure (CoP) between the legs during the stance phase. From a study performed by Langer, we see that pronation on the longer leg and supination on the shorter leg occurs as a compensatory mechanism for LLD [29]. Hence, we hypothesise that the CoP on the longer leg would be closer to the inner part of the foot, and on the shorter leg, it would be closer to the outer side. Therefore, analyzing the differences in CoP between both feet should be a good indicator of LLD.

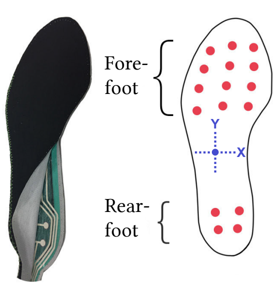

Figure 6: Layout of the pressure sensors (in red); 12 sensor positions are located at the forefoot and 4 spots are located at the rearfoot. Blue: calculated CoP.

Figure 6 depicts sensor locations; the blue sensor represents the centre of origin used for CoP calculations. The layout provides two groups, with front foot sensors, numbered 1-12, and rear foot sensors, numbered 13-16. The overall CoP at each time instance is calculated as a function of the mean pressure M_i and CoP$_i$ of each group, $i \in 1{,}2$:

$$CoP = \begin{pmatrix} CoPX \\ CoPY \end{pmatrix} = \frac{M_1 C_1 + M_2 C_2}{M_1 + M_2}$$

The CoP for each group, C_i and we mean, M_i, are calculated as shown by Elvitigala [16]. Particularly, CoP-X helps to observing the compensatory mechanism to LLD of pronation of the foot on the longer leg and supination of the foot on the shorter leg. We look at the CoP-X once the foot has ground contact. This information is provided by an increased Mean Pressure during gait cycle.

3.1.4 Feedback Augmentation. For a potential intervention, we deploy vibrotactile feedback, as this type of feedback is eyes-free and effective [2, 31, 32, 52]. Deploying vibration directly under the foot might create a relatable user experience for the wearer. For example, having an annoying pebble in the shoe between the foot and the insole is a known inconvenience, which can also change gait. We envision an augmented vibrotactile feedback to work in a similar way and to unconsciously alter gait. Therefore, we implemented eight Yuesui Coin Type Vibration Motors, which were embedded in a 3D printed UK size 10-11 buffer insole of 5mm thickness. These buffer insoles were 3D printed using Ninjaflex [39] filament, a thermoplastic polyurethane. Eight Sparkfun Haptic Motor Drivers (DRV2605L) connected to an Arduino were used to control all vibration motors. The application of augmenting vibrotactile feedback is achieved by mapping the CoP, calculated in real-time using Arduino code, to the vibration motors at the moment the foot touches the ground. The vibration is activated during Stance Phase to prevent irritating and numbing the foot. The motors vibrate between 200Hz and 400Hz, corresponding to its proximity to the CoP, where higher frequencies are applied to motors closest to the CoP. All vibration motors are driven at maximum power, and vibrotactile feedback insoles are placed on top of the pressure-sensitive insole to minimise signal attenuation through socks.

4 EVALUATION

The interest of this research is twofold. Firstly, we aim to explore detection capabilities of LLD with a fully mobile device, and secondly we aim to somewhat compensate an uneven gait caused by LLD, such as with augmenting vibrotactile feedback. We developed an artifact that potentially answers our research questions; however, this needs to be evaluated by the following user study.

4.1 Hypotheses

To answer our research questions, we established 5 hypotheses: When a person demonstrates LLD, we assume that...

- **H1** The person will walk in a curve to one side when blindfolded.
- **H2** The Stance Time will be different for each leg.
- **H3** The Ground Reaction Force will be different for each leg.
- **H4** The Centre of Pressure will be different for each leg.
- **H5** Augmenting vibrotactile feedback under the foot will alter gait.

4.2 Study Design

4.2.1 Apparatus. We used two insole prototypes (left and right foot) of the one described earlier. To assure an optimal fit, we inserted them into a blue pair of sports shoes. We used the same shoes for each person because the participants' shoes may have moulded to their feet and produced bias in our results. Although we tried to find participants with similar foot size, we taped the insole prototype to the shoe to reduce inaccuracies shifting may cause. We decided to induce a mild LLD by using an offset insole. This approach is a common method seen in several other clinical studies [5, 15, 33]. We 3D printed two insoles of 5mm and 10mm thickness *(see Figure 8)*. Preliminary trials found that the 10mm offset was a height that could fit comfortably within the shoe with the vibration prototype. The 10mm insole was placed inside the left shoe below the pressure-sensitive insole prototype to ensure the readings were not affected by the barrier that the 10mm insole would produce.

4.2.2 Participants. The conditions of the ethics committee required our study participants to have no known injuries or health conditions that could contribute towards existing gait asymmetry to qualify for the study. Thus, participants with known gait issues, such as osteoarthritis, skeletal deformities, or neurological conditions affecting balance [43], were excluded from the study. Participants were required to have a shoe size within 10.5 +/-1 (UK size). Sixteen people (6 male, 10 female) aged 18 − 23 years old (M = 20.4; SD = 1.54) participated in the study. Participants weighed between 45 − 84 kilograms (M = 63.9; SD = 9.48) and had heights ranging from 158 to 190 centimetres (M = 169.1; SD = 8.66).

4.2.3 Procedure. Participants attended an individual 60-minute session. To conduct the trials, we chose a wide (10m), long (20m), and flat (0° rise) space with no obstacles or difficult terrain. This ensured participants encountered no hazards while walking without visual perception. After participants were briefed on the trials and consent was given, they were fitted with the insole prototype. Once the participants were ready, the task containing several conditions started right away. Additionally, we had one person walk near the participant to assure their safety further, given the blindfold conditions would disable participants' "correction mechanisms." Having this support person complied with the university's ethics board requirements. We decided to blindfold the participant as it would improve the visibility of the LLD's impact on gait in our data. Aside from the blindfolds that switched off their visual sense, audio perception was also disabled. For this, we asked the participants to wear noise-cancelling headphones. When the participant was not blindfolded, we informed them to walk straight to the marked endpoint, which was 14m away. When the participant was blindfolded, we stopped them by tapping their shoulder (as they could not hear). When the participant was still standing, we marked down the ending position of their left and right legs and found the centre, to measure their walking deviation *(see Figure 7)*. We used stickers on the ground to mark how far the participant deviated from the origin and measured this using measuring tape. While blindfolded, the participant turned around, and we walked them back to the original starting position. We did this to ensure they did not know the magnitude of their deviation and bias the next two runs. To control as many conditions as possible, we artificially induced an LLD by an offset of 10mm using a 3D printed insole placed in the left shoe. The runs were completed three times for each trial.

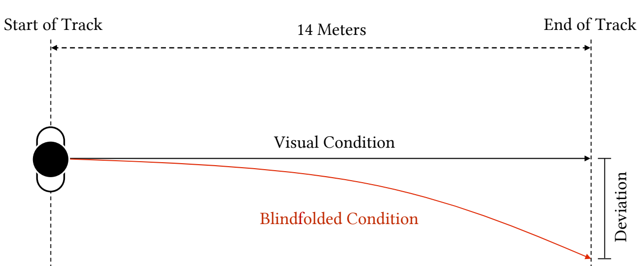

Figure 7: Schematic representation of the track (top view): Expectation of the Walking Deviation when a person is suffering from LLD when being blindfolded.

Figure 8: Image of the 10mm and 5mm insoles 3D printed from flexible filament (Ninjaflex [39]).

4.2.4 Task. Each participant completed eight conditions outlined in Table 1, with three trials (repetitions) of each condition performed. There were no further instruction than *"please try to walk 14 meters straight until we tap your shoulder"*. In the conversation with the participant, it was important to omit the fact that vibration could change their gait.

Table 1: Eight conditions were tested whereby several control conditions are also introduced.

Condition	Blindfolded	Induced Discrepancy	Vibrotactile Feedback
I	no	no	no
II	yes	no	no
III	no	yes	no
IV	yes	yes	no
V	yes	yes	left foot
VI	yes	no	left foot
VII	yes	yes	right foot
VIII	yes	no	right foot

4.2.5 Data Gathering. We compared the following dependent variables as measures to find whether there was a significance with how a person walked under each condition:

- Stance Time: Measured from both IMU and pressure sensor insole readings,
- Ground Reaction Force: Measured through the IMU readings,
- Centre of Pressure: Measured by the pressure sensors (CoP-Y shows gait cycle and CoP-X pronation),

Table 2: Clustered conditions in sets for data analysis.

Set	Conditions	Aim
1	I, II	For sets 1 & 2 we aimed to find out whether there is a natural LLD with the participant and
2	III, IV	whether we can observe any effect when blindfolding.
3	I, III	For sets 3 & 4, we compared the conditions with no offset verses 10mm offset. We aimed to find
4	II, IV	out if our induced offset is sufficient to detect the presence of LLD for future studies.
5	II, VI, VIII	For set 5 & 6, we aimed to find out about the effect of augmenting vibration. In set 5 we
6	IV, V, VII	applied vibration without offset and in set 6 when LLD was induced.

- Walking Deviation: Tape measured by manually recording how far a person deviated from the origin. Figure 7 shows the expected outcome when being blindfold when having an LLD.

For Stance Time, and GRF, and CoP we checked the differences in percent for each leg. For the experiments, IMU and pressure sensor readings were sampled at 100Hz, which was sufficient for recording walking data. The moment we actuated the vibration motors, the pressure sensors showed reading errors, and thus we only relied on IMU data during this condition.

4.3 Results

To analyse our data that will reveal whether we are able to identify LLD and influence gait, we had to run different comparisons between our conditions. We clustered the conditions in 6 sets that can be found in the following Table 2. Our data was treated as parametric data. Before selecting the statistical test, we conducted a Shapiro-Wilk test checking whether our assumption of normality is satisfied.

4.3.1 Set 1 & 2. For set 1, we compared if there is a significant difference for Stance Time, GRF, CoP, and Walking Deviation when participants had no offset. For set 2, we compared the same but with an induced LLD.

Stance Time: A paired sample t-Test for set 1 ($T(14)$ = -3.2708, $p < 0.01$) yielded a statistical difference between Stance Time with vision (M = 13.02; SD = 60.78) and blindfolding (M = -0.4; SD = 65.36). This result shows that disarming the visual correction mechanism has an impact on gait. This impact is also visible in set 2, where we ran the same test but with an induced LLD for both conditions. When blindfolded (M = -3.48; SD = 65.57) the person had a greater variance in Stance Time than with vision (M = 9.56; SD = 57.0) suggested by an F-Test ($F_{92,184}$ = 1.42; $p < 0.05$). The mean difference was close enough to show significance following a paired sample t-Test ($T(15)$ = -0.31968, p = 0.06).

Ground Reaction Force: A paired sample t-Test for set 1 ($T(14)$ = -2.1827, $p < 0.05$) yielded a statistical difference meaning that blindfolded walking has an impact on the gait as the GRF difference between legs changes between vision enabled (M = 22.0; SD = 104.83) and blindfolded (M=-2.51; SD = 93.42). In set 2, when having an LLD induced in both conditions: vision enabled (M = 17.03; SD = 92.13) and blindfolded (M = 1.04; SD = 89.91) the GRF difference is visible, although a pairwise t-Test ($T(15)$ = -0.27026, $p > 0.05$) did not yield a statistical difference due to the high variance in data.

Center of Pressure: For set 1, in terms of CoP-X, a pairwise t-Test did not identify any significant difference ($T(86)$ = -1.32; p = 0.1) between both conditions: enabled visual perception (M = 6.56; SD = 7.06) vs. blindfolded (M = 7.62; SD = 7.6). In terms of CoP-Y, a paired t-Test also could not identify a significant difference $T(175)$ = 1.84; p >0.05) in rollover movement between the visual condition (M = -25.5; SD = 33.8) and the blindfolded condition (M = -35.23; SD = 36.42). For set 2, we induced an artificial LLD, a pairwise t-Test ($T(175)$=-0.49; p >0.05) could not suggest CoP-X has a different pronation that might have occurred between the visual condition (M = 6.65; SD = 7.09) and the blindfold condition (M = 7.22; SD = 8.19). However, a pairwise t-Test identified significant differences in CoP-Y, which reflects a different rollover movement ($T(175)$=-0.49; p >0.05) between the visual condition (M = -25.65; SD = 33.82) and the blindfold condition (M = -37.23; SD = 35.21). This result is consistent with our other finding, blindfolding impacts gait.

Walking Deviation: For set 1, a pairwise t-Test did not show any statistical difference ($T(14)$ = 0.088115, $p > 0.05$) indicating that blindfolding the participant without any offset did not make them statistically deviate towards one side and end up at the target point with a mean deviation of 4.25cm. This result validates our assumption that the participants in our study did not have a significant LLD. For set 2, when we induced an LLD with the 10mm offset, a statistical difference was observed by a pairwise t-Test ($T(14)$ = 2.1952, $p < 0.05$). With induced LLD and without visual compensation mechanism, the participant walked in a curve towards one side as expected (*see Figure 7*). The participant ended up 74.62 SD = 206.94cm)away from the target point on average. Clearly, LLD is most prevalent in the blindfold condition.

4.3.2 Set 3 & 4. For set 3, we compared whether a significant difference with Stance Time, CoP, and GRF existed when we induced a 10mm LLD in comparison to our control condition. For set 4, we compared the same thing in blindfold. Here, we also look at the Walking Deviation.

Stance Time: The paired sample t-Test for set 3 ($T(14)$ = -1.58, $p > 0.05$) did not yield a statistical difference meaning the Stance Time is not different without LLD (M = 13.02; SD = 60.78) or with induced LLD (M = 9.56; SD = 57.0). This result evidences the human's visual compensation mechanism to have enacted. For set 4, when running the same test blindfolded, a paired sample t-Test ($T(15)$ = 0.18906, $p > 0.05$) also does not suggest a statistical difference between both conditions; no offset (M = -0.4; SD = 65.36) and induced LLD (M = -3.48; SD = 65.57). Thus, we conclude that Stance Time may not be a reliable method of detecting the presence of LLD.

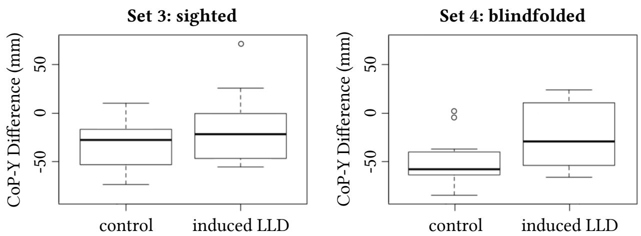

Figure 9: Left graph, set 3, shows how the CoP-Y, the foot's rollover movement changes when having an induced LLD. This effect becomes more significant when blindfolding the person, as depicted in the right graph, set 4.

Ground Reaction Force: A Shapiro-Wilk test for set 3 and 4 was done and found that the normality assumption was violated. Thus, we used a Wilcoxon signed-rank test for analysis. The Wilcoxon signed-rank test for set 3 when the participant was not blindfolded did not yield a statistical difference (V = 50, $p > 0.05$) between our control condition (M = 22.0; SD = 104.83) and the LLD condition (M = 17.03; SD = 92.13). With enabled visual perception, the participant did not exert more force on the ground from the heel contact point. The test for set 4 when the participant was blindfolded (V = 114, $p < 0.05$) had a statistical difference. When blindfolded the difference of GRF between both legs varies across both conditions the LLD condition (M = 1.04; SD = 89.91) in contrast to our control condition (M=-2.51; SD = 93.42).

Center of Pressure: For set 3, we conducted a paired sample t-Test to compare the differences in CoP-X between discrepancy groups. No significant difference ($T(14)$ = -0.101, p = 0.92) in the difference for CoP-X was found between the control condition (M = 6.72; SD = 5.55) and the LLD condition (M = 6.91; SD = 7.18) for sighted walking. When having a look at CoP-Y, *(see Figure 9)*, we can spot a change. However, a t-Test ($T(14)$ = -1.29, p = 0.22) could not evidence a significant difference for sighted walking, between the control condition (M = -30.9; SD = 26.8) and the LLD condition (M = -19.0; SD = 35.4). A greater offset than 10mm is likely to provide significance here. For set 4, we induced a discrepancy. A paired sample Wilcoxon signed-rank test also did not find significant difference in CoP-X (V = 73, p = 0.49). For CoP-Y a paired sample t-Test found a significant difference in CoP-Y ($T(14)$ = -3.82, p = 0.002) between the control condition (M = -48.4; SD = 27.8) and the LLD condition (M = -22.2; SD = 34.6) for blindfolded walking. With an induced LLD, the rollover movement of the foot differs, indicating that, on average, participants' mean CoP-Y during the stance phase is closer to the toe on their left foot compared to their right foot.

Walking Deviation: For set 3, no statistical analysis was performed. Our previous analysis underpins the visual correction mechanism to help the user to still walk in a straight line. On test 4, we performed a paired t-Test when the participants were blindfolded without (M = 4.25cm SD = 133.55cm) and with the 10mm offset (M = 74.62 SD = 206.94cm). There is a statistical difference ($T(15)$ = 2.4914, $p < 0.05$) exhibiting that while blindfolded, participants have a walking deviation. When inducing LLD the participants walk to one side *(see Figure 10)*, which confirms our previous finding.

4.3.3 *Set 5 & 6.*

For set 5, we wanted to compare if there was a difference in walking when augmenting vibrotactile feedback under the foot. We investigated this with applying feedback on the left and right foot. For set 6, we compared the same difference but with an induced LLD. Due to problems with our implementation, we were unable to read reliable data from the pressure sensors when actuating the vibrotactile feedback. Therefore, we are unable to calculate CoP.

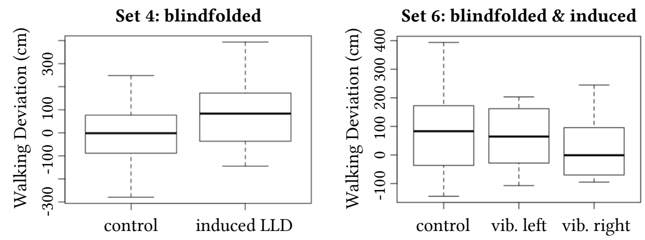

Figure 10: The left graph, set 4, shows the Walking Deviation from the center lane when the participant was blindfolded. Once LLD is induced on the left side, the participant tends to walk in a right curve ending up almost a meter away from the center. The right graph, set 6, shows how vibration on the right side enabled participants to correct the walked path towards the center.

Stance Time: For set 5, we applied a repeated measures ANOVA ($F_{2,45}$ = 7.716, $p < 0.01$) that yielded a statistical difference in the Stance Time *(see Figure 11 - left)*. When a participant had no vibration (M = -0.4; SD = 65.35) the difference of stance time between both legs were fairly similar. When vibration was present on the left (M = 21.1; SD=64.3), the Stance Time increased on the left foot. Likewise, vibration on the right side increased Stance Time for the right foot (M = -35; SD = 70.1). A post hoc analysis using Tukey's HSD revealed that the no vibration – right vibration and left vibration – right vibration pairs yielded a significant difference. The significance seems stronger for the right leg, which can be explained by its leg dominance. Therefore, stimulating the right foot might have a greater impact on gait parameters. Similar results can be found for set 6 where we induced LLD *(see Figure 11 - right)*. The repeated measures ANOVA ($F_{2,45}$ = 13.25, $p < 0.01$) also yield a statistical difference. The post hoc analysis with Tukey's HSD revealed a strong significance between no vibration (M = 3.48; SD = 65.57) and vibration on the right foot (M = -32.48; SD = 69.18). Vibrotactile feedback on the left foot (M = 18.44; SD = 66.58) also increased stance time; however, this was not significantly different to the control condition with no feedback. This result may be due to right leg dominance.

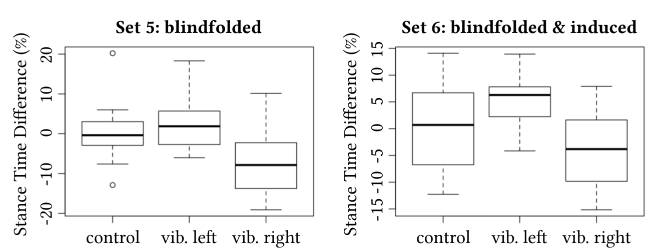

Figure 11: For both sets, the control condition shows no Stance Time Difference between both legs, although deviations occur. Augmenting a vibrational feedback under the foot shows a significant change in Stance Time. This alteration in gait can be used to compensate gait asymmetries caused by LLD.

Ground Reaction Force: The repeated measures ANOVA for set 5 ($F_{2,45}$ = 2.317 p > 0.05) did not yield a statistical difference in the GRF when the participant had no vibration (M=-2.51; SD = 93.42) and vibration under the left foot (M = 5.991; SD = 74.57) and under the right foot (M=-6.21; SD = 60.88). Therefore, we can conclude that vibration has no impact on the GRF. In set 6, we additionally induced LLD. The repeated measures ANOVA for set 6 ($F_{2,45}$ = 1.134, p > 0.05) also did not yield any statistical difference comparing the control condition without vibration (M = 1.04; SD = 89.91), vibration under the left foot (M = 5.19; SD = 73.29), and vibration under the right foot (M = -2.15; SD = 56.98). Thus, inducing vibration with and without offset does not affect the GRF exerted from the heel to ground contact for either leg.

Walking Deviation: For set 5, we compared three conditions: when the participant had no vibration (M = 4.26cm; SD = 133.55cm) - slightly deviating to the right due to right leg dominance, vibration at the left foot (M = 32.25cm; SD = 126.48cm) - making them deviate even more towards the right, and vibration at the right (M = -4.2cm; SD = 171.83cm) foot - making the participants to deviate to the left. As we did not control leg dominance, great deviation occurred in our data, which did not allow a repeated measures ANOVA ($F_{2,45}$ = 0.221, p > 0.05) to identify a statistical significance. In set 6, we compared the same thing but with induced LLD. With no feedback, the participants strongly deviated to the right by 74.62cm (SD = 206.94cm) from the center position. Also, with vibration on the left foot, participants deviated strongly to the right by 58.0cm (SD = 134.31cm). When vibration was present on the right foot, the route could be corrected and the deviation decreased, resulting on only 24.08cm (SD = 129.02cm) walking deviation. However, the repeated measures ANOVA ($F_{2,45}$ = 0.802, p > 0.05) was unable to identify statistical difference due to the high variance.

4.4 Answering Hypotheses

To answer both of our research questions, we established five assumptions. When a person demonstrates LLD:

The person walks in a curve to one side when blindfolded. We can accept hypothesis **H1**. All of our conditions unambiguously verified that when disarming the visual compensation mechanisms by blindfolding, a person with LLD will walk in a curve.

The Stance Time is not different for each leg. We reject hypothesis **H2**. We found that the difference in Stance Time was not statistically significant with a 10mm offset. We conclude that Stance Time may not be a reliable method to detect LLD. However, a tremendous LLD could show differences, which remains unknown for us.

Under certain conditions, the Ground Reaction Force can be different for each leg. We neither accept nor reject hypothesis **H3**. Our analysis shows that there is a significant difference in the GRF when the participant had an induced LLD when blindfolded.

Under certain conditions, the Centre of Pressure can be different for each leg. We take a cautionary approach and neither accept nor reject **H4**. When the visual compensation mechanism is disarmed, LLD indeed impacts on the rollover movement of the foot, which is reflected at the y-axis of the CoP. There were also differences at the x-axis of the CoP, which reflects pronation and supination. However, this finding was not statistically significant.

Augmenting vibrotactile feedback under the foot alters gait. We can confirm hypothesis **H5**. Augmenting vibrotactile feedback at the foot impacted our measured parameters. In particular, the difference in Stance Time between both legs increased when triggering vibrations during the foot's ground-contact time. Other parameters also changed; however, they were not statistically strong enough, which may be the result of a rather limited sample size. Increasing the feedback's intensity might further increase the effect.

5 DISCUSSION

5.1 Summary

We were interested in investigating whether the between leg difference in Stance Time, GRF, and CoP calculated from plantar pressure and IMU data and blindfolded deviation could be used to infer an LLD. To reduce complexity of the study, we only induced LLD on the left side. We could see this impacted the foot's rollover movement and thus the user tended to walk in a curve to the right side when blindfolded. We were then interested in whether vibrotactile feedback could alter the asymmetry of gait. We found that vibration on the right foot can alter gait as it slightly increases the Stance Time at the right foot, which is done unconsciously by the user. This corrected the curved walking deviation more towards the initial center lane. We rate this as promising results and discuss our key insights, limitations & challenges, and future work.

5.2 Key Insights

Pronation remains unchanged. We believed that pronation and supination due to LLD would be evident in the centre of pressure data on the x-direction. With a longer left leg, we hoped to see a CoP closer to the inside of the left foot and the opposite for the right foot, giving a more positive CoP difference value. Even if true, the sensor readings were not fine-grained enough to notice this.

Rollover movement changes. Only CoP in the y-direction while blindfolded was found to have a significant difference between discrepancy at LLD. We saw that, on average, the mean difference in CoP-Y when blindfolded without a discrepancy is between 11.5mm and 40.8 mm less than with the induced discrepancy. When having an offset at the left side, the average participants' mean CoP-Y during stance phase is closer to the toe on their left foot compared to on their right foot.

Visual correction mechanism. Like CoP, the GRF difference between legs was found only to have a significant difference between discrepancies when the user is blindfolded. It indicates that visual perception works as a great correction mechanism to aid in gait symmetry.

Using blindfolds as a useful measure. To investigate the impact of augmenting vibrotactile feedback on the symmetry of a participant's gait, the blindfolded deviation was useful as a measure to detect LLD. The results found that, on average, a participant deviated more to the right when they had a 10mm longer left leg than they did with no discrepancy. Although we only focused on a 10mm discrepancy on one side, we believe this can be extended to a discrepancy on the other leg, and that a correlation between deviation and LLD size may exist.

High walking deviation with blindfolds. It was noticeable that blindfolding, even without inducing LLD, introduces some deviation from the straight path. This is due to reduction of the person's sense of balance, which is also supported by visual perception. Another impact are postural influences and other random extrinsic factors.

Impacting gait parameters with vibrotactile feedback. Applying vibrotactile feedback to either the left or right foot did not result in a significant difference in blindfolded Walking Deviation compared to when no feedback is applied with either discrepancy. This is also due to the high standard deviation across users. GRF is not significantly deviated. However, the difference in Stance Time is significantly increased when augmenting vibrotactile feedback.

Altering vibrotactile feedback style. In this study, the vibration was applied once the foot hit the ground and then closest to the centre of pressure. Other methods of applying vibration could influence gait also, such as applying a constant vibration, increasing strength of stimulus, or introducing specific vibration patterns.

Implicit Behaviour Change. Another interesting aspect is that although the user notices feedback, the change in gait happens without the person to consciously think of a behaviour change.

5.3 Limitations & Challenges

The nature of prototyping. As we are not using a commercial product, our prototype is fragile, prone to breakage, and needs frequent maintenance. Missing or corrupted data is natural. We had to exclude some data from participant 7 and participant 9. Moreover, reliable CoP data could not be calculated during actuating vibration motors, which is unfortunate.

Current detection difficulties. When we did not blindfold the user, differences in Stance Time, GRF, and CoP in other conditions did not show significant differences between discrepancies. The lack of evidence for detecting LLD from these features could be due to the quality of the calculated measures. Optimising function parameters to individual participants or individual data sets could improve the feature calculations for Stance Time and CoP by better identifying ground contact points and the pre-swing phase. Also, a higher-grained pressure-sensing insole could be beneficial. Moreover greater LLD than 10mm are envisioned to be easier detectable.

Leg Dominance. Studies [4, 50] show that ~85% of the population might be right leg dominant. In assuming that leg dominance would not greatly impact our data, we neglected to control this variable. However, after running our experiments, we observed this factor to have a greater impact than initially expected. From our data, at least 25% (4 participants) seem to be left leg dominant, which resulted in greater noise. Future work should consider this factor.

Limited resources. For our study, we were limited by many resources. For instance, finding a suitable space that is long, wide, and flat enough without obstacles proved difficult. Our track had a length of 14 metres only. User study sessions were scheduled for 60 minutes due to the availability of participants. This time limitation restricted the number of runs per experiment that could be collected. Participants often had to come in for a second session.

Practicability. Our studies were carried out in a controlled environment, which differs from reality. Environmental context, such as ground slope and surface, can also influence the way the user walks, which impacts the data collected. In our experiment, we determined a straight path to follow. In a product-ready device, other sensors might be required to help obtain information on the nature of the track, such as GPS.

Generalizability. A temporary FLLD could even create greater discrepancies in gait than a 10mm offset that we used. Also, we only included young healthy people. Other age groups and people in different health conditions, namely those suffering from actual LLD, could be included. LLD could be pronounced differently among these groups, although we expect similar results.

5.4 Future Work

External Validity. The aim of this research was to "Construct Validity" [10] of using a new method and device. Evidencing “External Validity” (either in stages of 1. “Robustness”, 2. “Ecological Validity”, or 3. “Relevance”) would be the next goal in this research. Particularly, creating “Ecological Validity” requires representative studies that extend to everyday life conditions and to a large population in the world [10]. The next paragraphs outline the further steps towards this goal.

Machine Learning. To detect LLD, we calculated several features from pressure sensor data and IMU, such as Stance Time, GRF, and CoP. To better exploit the rich source of information available from time-series data, supervised machine learning approaches could be explored. We expect machine learning to provide highly accurate results. A similar approach to a study by Matthies et al., which identified users and ground surfaces using smart insoles and machine learning algorithms [34], could be investigated for identifying LLD.

Unobtrusive Design. The device is fairly simple and only requires an IMU, a pressure sensitive insole, a few vibration motors, a microcontroller, and a battery. For convenience, we used rather bulky prototyping equipment, such as an Arduino, which is hidden in a 3D printed case strapped to the ankle. However, the entire device has great potential to be unobtrusively integrated into a shoe.

Real-Time Detection. With an improved sensing device integrated in a shoe, capable of immediately detecting LLD on-the-go, applying vibrotactile feedback would only actuate when an uneven gait is detected. This would enable overcoming negative health-related consequences caused by LLD. For instance, a muscle spasm after an exhausting football training would be detectable and compensated by an automated feedback augmentation in the required intensity.

6 CONCLUSION

In this project, we explored the detection of mild LLD with an insole-based approach using pressure-sensors and IMU. We found that augmenting vibrotactile feedback under the foot could influence gait, as there was a substantial difference in Stance Time. Although not yet statistically significant due to limitations in our study setup, vibrotactile feedback could correct the Walking Deviation from a strong curve back towards the center lane when the participant is blindfolded. Our results conclude that in the future, a well-calibrated system, such as one implemented in a smart shoe, may significantly contribute to a more symmetrical gait and help reduce the adverse effects leg length discrepancies can cause, including remedying back pain and headache.

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