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Biomechanical analysis of walking gait when simulating the use of an Ilizarov external fixator.

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Keywords: Ilizarov frame, Gait Analysis, Biomechanics, Kinetic, Kinematics

Abstract

The Ilizarov frame is an external fixation device, primarily used for the treatment of complex fractures. The authors postulate that the size and weight of the frame may lead to biomechanical adaptations to gait, independent to any injury.

Temporospatial characteristics, kinetics and kinematics were assessed when simulating the use of an Ilizarov frame. Fifteen healthy participants performed walking trials, with and without the simulated frame.

Significant changes to temporospatial characteristics were identified, with a decreased mean walking speed (with: 1.24 $m \cdot s^{-1}$; without: 1.29 $m \cdot s^{-1}$) and increased mean step width (with: 0.14 m; without: 0.11 m). The push-off phase of gait differed significantly between test conditions with mean increases in ankle dorsiflexion angles (with: 90.4°; without: 89.0°) and extension moments (proportional to body weight or P BWT) at the knee and ankle (Knee with: 0.8 P BWT·m; without: 0.7 P BWT·m; Ankle with: 1.6 P BWT·m; without: 1.6 P BWT·m).

Although changes were small and likely to be clinically insignificant, the size and weight of the frame led to adaptations which may be magnified for patient groups with associated injury and pain at the lower limb. Results provide an argument for the potential redesign of the frame.

Key Words: motion capture, kinetic, kinematic

Word Count: 3940

Introduction

External fixation is a surgical intervention in which bone is stabilised at a distance from a fracture site to promote healing and / or correct deformity. In lower limb fractures, external fixation is the definitive surgical method employed when severe damage to soft tissues has occurred[1]. Over a million cases of external fixation can be expected to take place across the world per year[2]. Several external fixation devices are commercially available, one of which is the Ilizarov frame, a modular system allowing constructs to be tailored to specific situations. Ilizarov frames are comprised of relatively rigid metal rings connected by threaded rods, with each ring attached to bone by comparatively flexible pins or wires (Figure 1). The use of relatively flexible pins and wires allows some load to be transferred to the bone on weight bearing, resulting in potentially advantageous mechanical stimulation of healing bone[3]. During weight bearing, tensioned wires hold the fractured bone in the required alignment whilst allowing for axial movement at the fracture site. This axial micro-motion has been demonstrated to enhance callus formation and ultimately bone healing[4].

Tibial fracture accounts for the majority of all long-bone fractures and is routinely treated with external fixation when complex[1,5]. Tibial fracture patients treated with Ilizarov frames display abnormal walking patterns for a variety of reasons. Gait adaptations such as abnormal kinematics at the hip and knee (described as antalgic strategies), asymmetry between loading patterns and alterations to temporospatial characteristics have been identified in individuals with external fixation of the lower limb[6,7]. Joint stiffness, laxity of the knee and transient foot drop has also been reported in patients treated by definitive external fixation[8-10]. Alterations to an individual's natural joint kinetics during treatment may lead to loss of function and abnormal loading of the fractured limb. Avoidance of the affected limb may adversely impact joint and muscle conditioning, as well as potentially slowing bone healing due to a lack of axial micro-motion at the fracture site[4]. From the current literature, it cannot be concluded whether the biomechanical abnormalities reported are influenced by the fracture, the external fixation device, or a combination of the two.

Wearing an Ilizarov frame in itself will potentially cause gait abnormalities. The ring diameter will vary with the anatomy of the patient, with a minimum gap of approximately 2 cm between the limb and the ring recommended to accommodate swelling and allow soft tissue care[11]. The usual diameter of rings used in adult patients is between 140 to 180 mm and they will frequently protrude 40 to 50 mm on the medial side of a patient's leg. Four 160 mm stainless steel rings with interconnecting rods weighs approximately 1.5 kg without the additional hardware to connect the frame to bone. Many frames will weigh considerably more than this. The addition of bulk and weight to the lower limb, when wearing the frame, will increase the inertia at the limb and is therefore likely be expressed through adaptations in biomechanical data. The impact of the bulk and weight of the frame itself on gait, independent of a patient’s injury, has not been determined experimentally. Without understanding this potential impact, it is difficult to explore rationales for the innovation and development of new fixation devices which may ultimately benefit the patient.

The authors thus postulate that wearing an Ilizarov frame, independent to injury, will lead to significant gait adaptations which may be detrimental to the loading of the affected limb and therefore fracture healing. The identification of biomechanical adaptations, due to the frame, may suggest grounds for the re-design of the frame. The global aim for this study was to investigate kinetic and kinematic changes to an individual’s gait when simulating the use of an Ilizarov frame, to determine the contribution of the bulk and weight of the frame itself. This aim was met through identifying biomechanical differences between normal walking and walking with the simulated Ilizarov frame. The use of healthy subjects allowed for the isolation of the impact of frame, independent to any injury. Specific objectives included determining differences between test conditions for: 1) vertical ground reaction force, 2) kinematics at the ankle, knee and hip, 3) joint moments at the ankle and knee. The author hypothesises that: 1) the size of the frame will increase stride width, therefore altering lower limb kinematics; 2) the weight of the frame will increase loading at the lower limb.

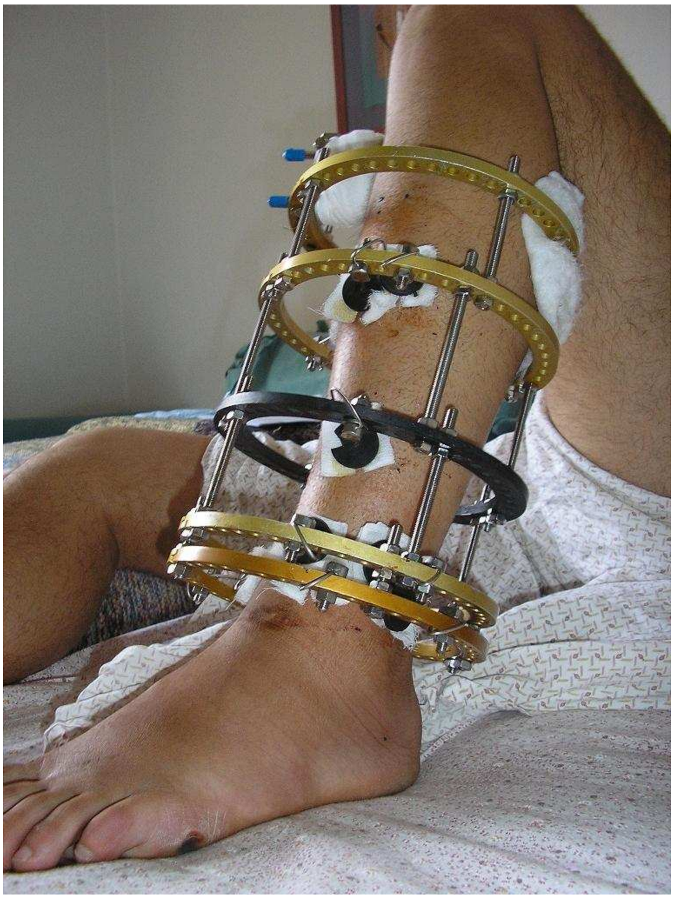

Figure 1. Ilizarov frame treating a fracture to the left tibia (Viapastrengo, 2007) (Wikimedia Commons).

Methods

Subjects: Twelve male and three female individuals were recruited from staff and students at the University of Leeds (Mean and standard deviation. Age: 22.5 y (SD ±3.1); Height: 1.79 m (SD ±0.09); Body mass: 73.7 kg (SD ±14.1)). Subjects were healthy and free from any injury, illness or pathology that could impact their natural gait. This ensured that results reflected the impact of the frame, independent to injury. Before data collection, written informed consent was obtained and a screening questionnaire was completed by all subjects. A risk assessment form was completed prior to the study and the protocol was ethically approved according to the guidelines of The University of Leeds ethics committee (BIOSCI 14-013).

Test conditions: Participants completed two test conditions at a self-selected, comfortable walking speed in their everyday footwear. The first test condition was a normal walk, used to identify baseline gait characteristics (Control condition or 'CC'). The second test condition involved the simulated attachment of an Ilizarov frame below the knee of the dominant leg (Simulated frame condition or 'SFC'). The Ilizarov frame included four rings, each with a diameter of 21 cm and weighed 1.23 kg in total. For the SFC the device was attached around the shank of the dominant leg, with a high density foam sheet fastened between the leg and the frame to ensure a secure attachment (Figure 2). Subjects walked across a fifteen meter walkway, reaching the first force platform in six steps. For both test conditions, subjects stepped on the first force platform with their dominant foot and the second with their non-dominant foot and repeated the activity sufficient times to obtain ten usable data sets per condition.

Experimental set-up: For each trial, twenty-eight 15.9 mm pearl retro reflective markers were attached to lower limb anatomical landmarks, in accordance with Visual 3D marker set guidelines[12]. Landmarks were identified through manual palpation in accordance to standardised techniques[13]. Additionally, tracking markers were attached to the thigh and shank, using four marker semi-rigid thermoplastic shells. Sixteen markers were attached to the frame itself, equalling a total of fifty-six markers for the whole body during the frame trials (Figure 2).

Data collection: Kinematic data was collected using a thirteen-camera Qualysis Oqus 3-D motion capture system at a frequency of 400 Hz (Qualisys™ Medical AB, Goteborg, Sweden). Ground reaction forces (GRF) were collected using two, in line, 600 x 400 mm AMTI (BP400600) embedded force platforms (AMTI, Advanced Mechanical Technology Inc., Watertown, MA, USA), synchronised to the camera system and sampled at 1200 Hz. A static trial was completed for each subject prior to the collection of dynamic trials for both the CC and SFC, to allow anatomical marker positions to be identified.

Data processing: Kinematic targets were filtered at 10 Hz using a Butterworth low pass filter[14,15]. The 'V3D_Composite_Pelvis' method was used to model the pelvis segment. Left and right iliac crest and sacrum markers were defined to achieve this. This allowed the thigh segment to be defined using the hip joint centre as the proximal joint, with the medial and lateral knee markers defining the distal joint. The shank segment was defined using medial and lateral knee and ankle markers, whereas the foot segment incorporated ankle (n=2), calcaneus (n=3) and metatarsal (n=4) markers (Visual 3D standard, v5.01.18, C-Motion, Germantown, MD, USA). Virtual landmarks were defined from the frame markers and located at the centre of each ring. Each ring was then modelled using anterior, posterior, medial and lateral markers and assigned a weight of 0.308 kg each. Joint angles for the ankle (flexion-extension), knee and hip (flexion-extension and abduction-adduction) (°) were defined through the orientation of one segment in relation to another. Internal joint moments (Proportional to body weight or

P BWT·m) for the ankle (flexion-extension) and knee (flexion-extension and abduction-adduction) were determined using a Newton-Euler inverse dynamic calculation. Both angles and moments were resolved into the proximal segment coordinate system. Moments were normalised to body weight and accounted for the weight of the frame. Speed ($m \cdot s^{-1}$), stride width (m), stride length (m), dominant step length (m) and non-dominant step length (m) were also calculated. These variables were investigated because pilot work indicated potential variation between conditions.

Statistics: Descriptive statistics were completed for data sets (means and standard deviations). Angular and joint moment data sets were found to be normally distributed through Shapiro-Wilk tests ($P \leq 0.05$). Data peaks were averaged for participants and paired T-tests calculated significance between data sets ($P \leq 0.05$). Effect size was calculated alongside any paired T-tests[16] (SPSS, v22.0, IBM Corp, Armonk, NY).

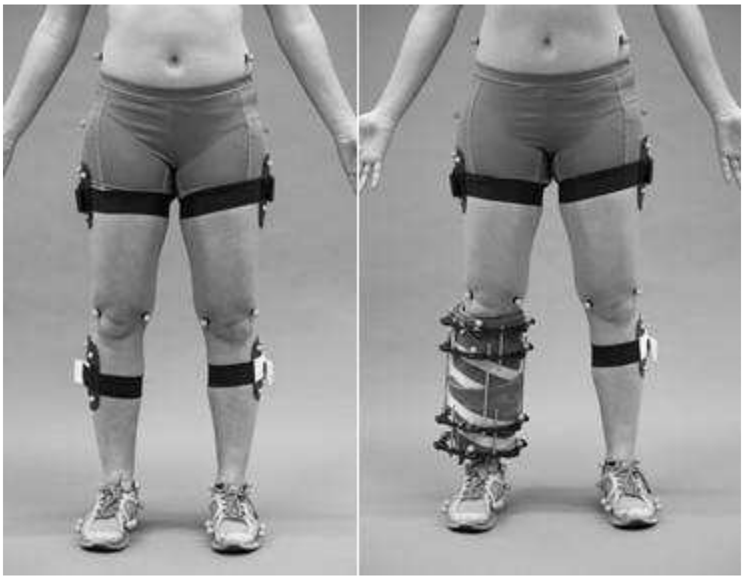

Figure 2. Anterior view of subject with marker setup for the control condition (left) (n=44) and simulated frame condition (right) (n=56).

Results

Mean walking speed was higher and significantly different for the control condition (CC) compared to the simulated frame condition (SFC) (CC = 1.29 m·s^{-1} (SD 0.12); SFC = 1.24 m·s^{-1} (SD 0.13)) ($P \leq 0.05$) (Table 1 and 2). Mean stride width increased when wearing the simulated frame (CC = 0.11 m (SD 0.02); SFC = 0.14 m (SD 0.02)) (Table 1) and was significantly different ($P \leq 0.05$) with a small effect size (Table 1). Although the mean stride length decreased with the simulated Ilizarov frame (CC = 1.38 m; SFC = 1.32 m) (Table 1), the data was not statistically significant ($P \geq 0.05$) (Table 1). No significant differences were identified for the non-dominant step length between test conditions or between the dominant and non-dominant leg for either condition ($P \geq 0.05$). However, the dominant leg (with the frame attached) showed a small decrease in mean step length from the CC (0.74 m) to the SFC (0.73 m). This small mean difference was significant ($P \leq 0.05$), with a medium effect size (Table 1).

The two GRF peaks (loading and propulsion, respectively) were similar for the CC (1.11 (SD 0.07) P BWT and 1.12 (SD 0.07) P BWT) and the SFC (1.12 (SD 0.07) P BWT and 1.13 (SD 0.07) P BWT) (Figure 3). The peaks seen in the data (GRF_1 and GRF_2) were each compared between test conditions (Table 1).

The mean ankle plantarflexion-dorsiflexion angle followed similar trends for both test conditions (Figure 4). The first plantarflexion peak (Ankle_Angle_X_P_1) did not show a significant difference between test conditions ($P \geq 0.05$). The dorsiflexion peak (Ankle_Angle_D) and the second plantarflexion peak (Ankle_Angle_X_P_2) showed mean increases when wearing the frame and were significantly different between test conditions ($P \leq 0.05$), with large and medium effect sizes, respectively (Table 1 and 2).

A significant difference was not identified between test conditions for the peak knee flexion angle ($P \geq 0.05$) (Table 1 and 2). Mean knee abduction-adduction data indicated less net adduction movement for the SFC than the CC (Figure 4). During the swing phase, the abduction peak (Knee_Angle_Y_Ab) was larger for the SFC, when compared to the CC (CC = -1.5° (SD 5.2); SFC = 2.1° (SD 4.9)). The negative value for the CC indicates a net adduction angle, highlighting a larger mean abduction angle experienced by the SFC at this point. This peak was significantly different between test conditions ($P \leq 0.05$) with a large effect size (Table 1). The adduction peak following this (Knee_Angle_Y_Ad) showed a smaller adduction value for the SFC than the CC (CC = -8.3° (SD 5.2); SFC = -5.7° (SD 4.1)). Again, this peak was found to be significantly different between test conditions (P ≤ 0.05), with a large effect size (Table 1). The increased magnitude of the abduction angle and the reduced adduction angle when wearing the simulated frame indicates a net increase in abduction at the knee.

The hip flexion-extension angle showed significantly different peak extension angles between test conditions (P ≤ 0.05) (Table 1, Table 1 and Figure 4). However, the standard deviation for the hip flexion-extension angle (CC = ±8.3°; SFC = ±8.4°), suggested increased variability when compared to the knee and ankle. The medium effect size further suggests overlap between the two data groups. The hip abduction-adduction angle showed similar trends for both test conditions, with a lack of significance between peak data (P ≥ 0.05) (Table 1).

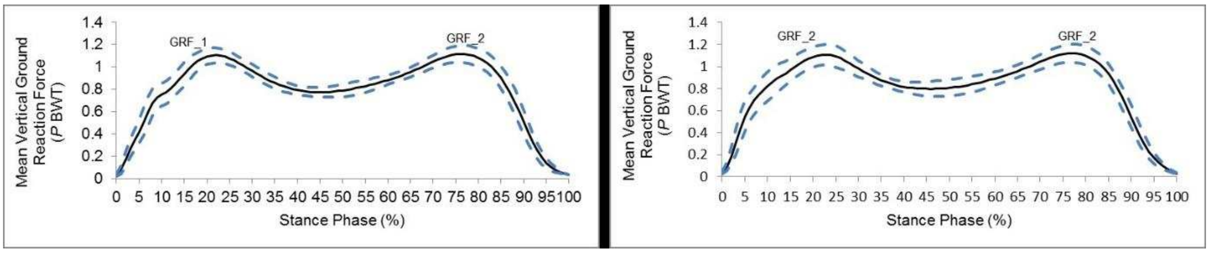

Figure 3. Mean vertical ground reaction force (proportion of body weight) during the stance phase of gait for a normal walk without (left) and with (right) the simulated Ilizarov frame. One standard deviation above and below the mean are shown as blue dashed lines. Peaks of interest are labelled, with an asterisk representing a statistically significant difference between the normal walk and walking with a simulated Ilizarov frame.

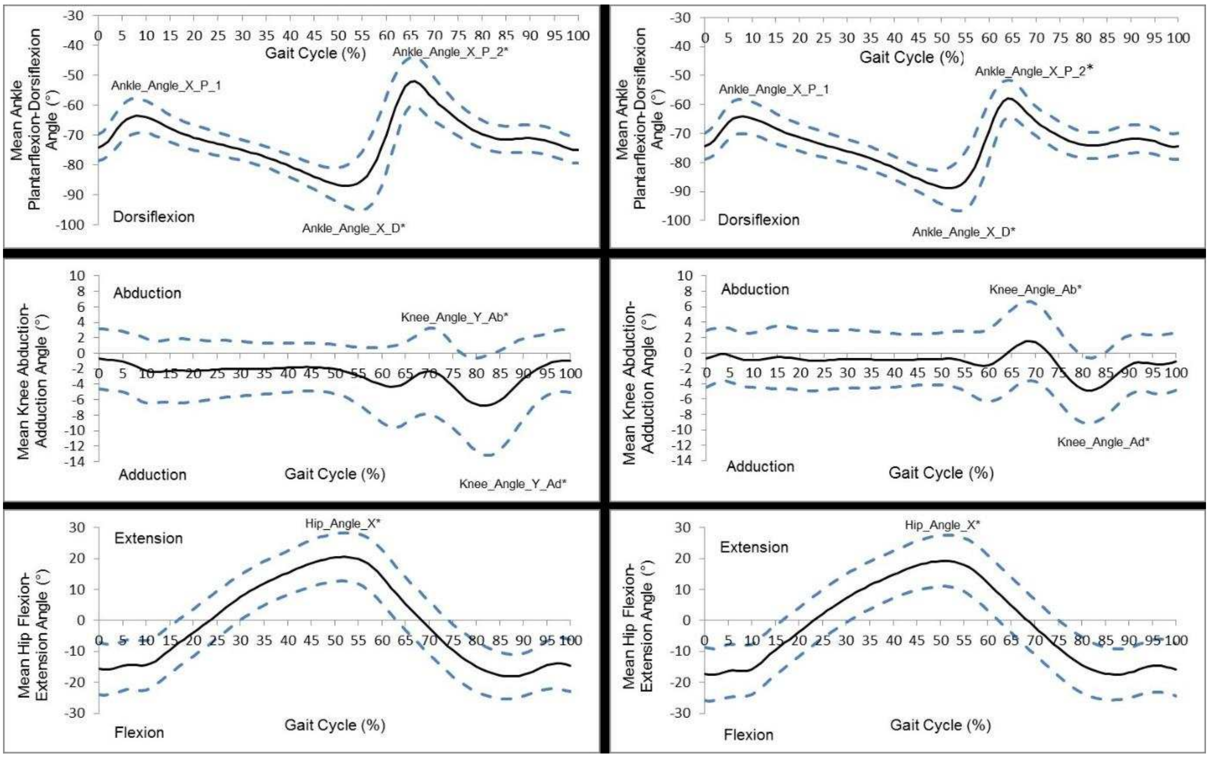

Figure 4. Mean angular data without (left) and with (right) the simulated Ilizarov frame for ankle plantarflexion-dorsiflexion (top), knee abduction-adduction (middle) and hip flexion-extension (bottom)

during one gait cycle. Standard deviation above and below the mean are shown as blue dashed lines. Peaks of interest are labelled, with an asterisk representing a statistically significant difference between the normal walk and walking with a simulated Ilizarov frame.

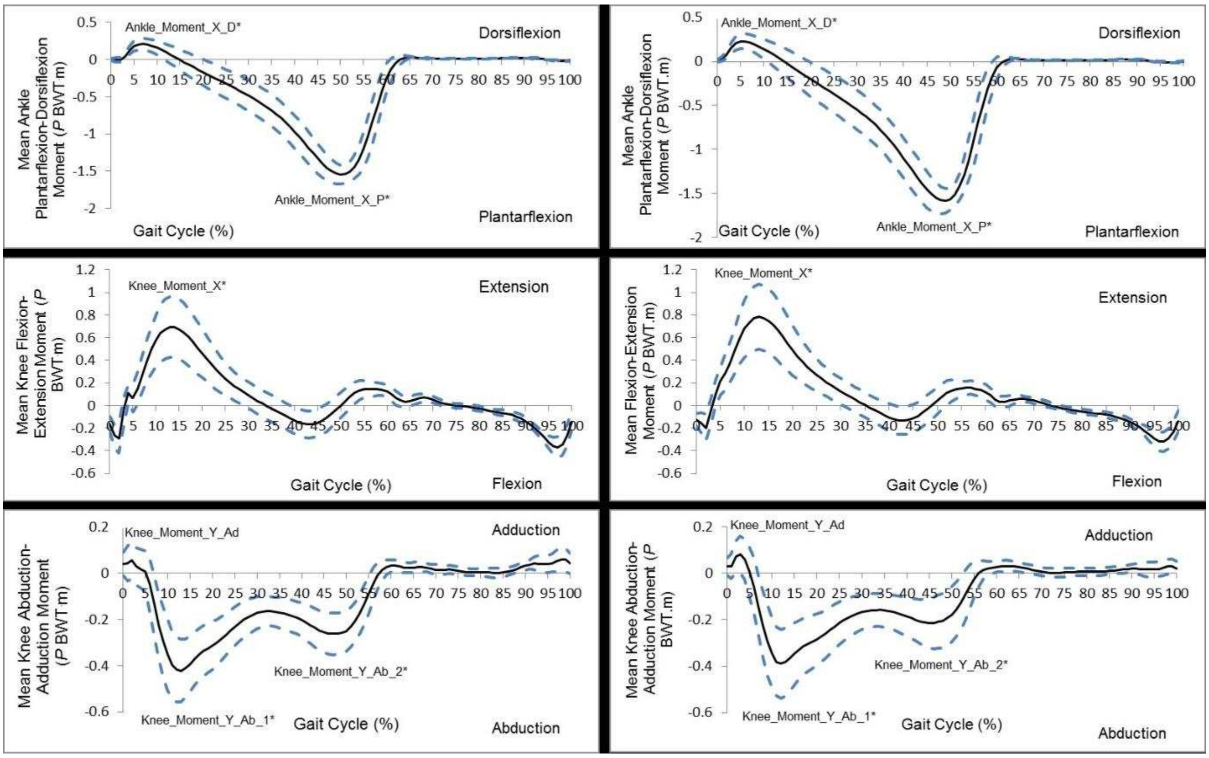

Figure 5. Mean moment data (proportion of body weight) without (left) and with (right) the simulated Ilizarov frame for ankle plantarflexion-dorsiflexion (top), knee flexion-extension (middle) and knee abduction-adduction (bottom) during one gait cycle. Standard deviation above and below the mean are shown as blue dashed lines. Peaks of interest are labelled, with an asterisk representing a statistically significant difference between the normal walk and walking with a simulated Ilizarov frame.

Table 1. Mean temporal distance calculations, peak ground reaction force (GRF), peak joint angles and peak joint moments with standard deviations for the control walk and for walking with the simulated Ilizarov frame. (Peaks relate to points of interest seen in the data with an asterisk representing a significant difference). (X=Flexion-Extension Y=Abduction-Adduction)

Temporal Distance Calculation	**Walk Mean**	**Standard Deviation (±)**	**Frame Mean**	**Standard Deviation (±)**	**P(T<=t) two-tail**	**Significant P≤0.05**	**Effect Size (S,M,L)**
Speed (ms^{-1})*	1.29	0.12	1.24	0.13	0.00	✓	0.60 (L)
Stride Width (m)*	0.11	0.02	0.14	0.02	0.00	✓	1.25 (L)
Stride Length (m)	1.38	0.36	1.32	0.45	0.44	×	0.19 (S)
Dominant Step Length (m)*	0.74	0.06	0.73	0.13	0.02	✓	0.19 (S)
Non-Dominant Step Length (m)	0.72	0.05	0.72	0.05	0.57	×	0.22(M)
Peak GRF	**Walk Mean (P BWT)**	**Standard Deviation (±)**	**Frame Mean (P BWT)**	**Standard Deviation (±)**	**P(T<=t) two-tail**	**Significant P≤0.05**	**Effect Size**
GRF_1	1.11	0.07	1.12	0.08	0.15	×	0.20 (S)

GRF_2	1.12	0.07	1.13	0.08	0.14	×	0.03 (S)
Peak Angle	**Walk Mean (°)**	**Standard Deviation (±)**	**Frame Mean (°)**	**Standard Deviation (±)**	**P(T<=t) two-tail**	**Significant P≤0.05**	**Effect size (S,M,L)**
Ankle_Angle_X_P_1	-63.3	5.5	-63.4	5.8	0.94	×	0.01 (S)
Ankle_Angle_X_P_2*	-50.4	8.8	-56.1	5.5	0.01	✓	1.10 (L)
Ankle_Angle_X_D*	-89.0	7.1	-90.4	7.1	0.00	✓	0.29(M)
Knee_Angle_X	73.6	4.9	72.4	5.6	0.07	×	0.33 (L)
Knee_Angle Y_Ab*	-1.5	5.2	2.1	4.9	0.00	✓	1.00 (L)
Knee_Angle Y_Ad*	-8.3	5.2	-5.7	4.1	0.00	✓	0.79 (L)
Hip_Angle_X*	20.9	8.3	19.5	8.4	0.00	✓	0.24(M)
Hip_Angle_Y	6.4	2.8	6.0	3.1	0.36	×	0.21(M)
Peak Moment	**Walk Mean (P BWT·m)**	**Standard Deviation (±)**	**Frame Mean (P BWT·m)**	**Standard Deviation (±)**	**P(T<=t) two-tail**	**Significant P≤0.05**	**Effect size (S,M,L)**
Ankle_Moment_X_D*	0.2	0.1	0.2	0.1	0.00	✓	0.50 (L)
Ankle_Moment_X_P*	-1.6	0.1	-1.6	0.1	0.05	✓	0.38(M)
Knee_Moment_X*	0.7	0.3	0.8	0.3	0.01	✓	0.46(M)
Knee_Moment_Y_Ad	0.1	0.1	0.1	0.1	0.11	×	0.28(M)
Knee_Moment_Y_Ab_1*	-0.4	0.1	-0.4	0.1	0.05	✓	0.38(M)
Knee_Moment_Y_Ab_2*	-0.3	0.1	-0.2	0.1	0.00	✓	0.59 (L)

Discussion

Significant differences were identified between a number of temporospatial, kinetic and kinematic variables when wearing the simulated Ilizarov frame compared to without. It is likely that these occurred due to the added size and weight at the lower limb.

Temporospatial calculations: The small but significantly slower walking speed for the simulated frame condition (SFC) (1.24 $m \cdot s^{-1}$) than the control condition (CC) (1.29 $m \cdot s^{-1}$) could be attributed to the added weight (+1.23 kg) and therefore increased inertia at the lower limb when wearing the frame, but may also be a compensation for unfamiliarity of walking with the device attached. As would be expected, the decreased walking speed whilst wearing the frame was identified alongside a significant decrease in the dominant step length (the frame side). This further suggests that it is the frame which leads to the alterations, as the non-dominant leg step length showed no significant differences between test conditions. Previous literature identified a significantly reduced step length when treated with an external fixator[7], increased asymmetry between limbs when wearing a unilateral ankle weight[17] and an increased risk of tripping when stepping over obstacles whilst wearing heavy duty boots[18]. These findings support the suggestion that the increased inertia at the dominant leg, due to the Ilizarov frame, may be the cause of alterations to step length.

Stride width was also significantly increased for the SFC. It is reasonable to assume that this resulted from the size of the frame (diameter: 21 cm), rather than the weight. This adaptation may be essential to ensure efficient gait with the frame. The increased stride width would be expected to have shown kinematic changes at the hip (increased abduction), in order to alter the position of the dominant leg. This was not the case. As the mean change is relatively low between the two conditions (CC = 0.11 m; SFC = 0.12 m), rather than a clear alteration in the hip abduction angle, the increased stride width may have been due to an accumulation of a number of small kinematic changes.

Ground reaction force: The first peak seen for the GRF data represented the point of weight acceptance, whereas the second peak was the propulsive phase[19,20]. Vertical GRF data showed similar trends for both conditions and was comparable to previous findings for healthy subjects[19,21,22]. It was hypothesised that the SFC would increase loading at the lower limb. This statement must be rejected. However, it is possible that the loading axes of the limb are altered through adaptations to angular and temporospatial characteristics. This may therefore influence joint moments, irrespective of the unchanged vertical ground reaction force.

Joint moments and motions: The peak dorsiflexion angle, occurring at approximately 50% of the gait cycle, represents the point at which the plantarflexors are at peak contraction, in order to propel the body forwards[20,23,24]. The mean peak dorsiflexion angle (Ankle_Angle_X_D) occurred at approximately 50% of the cycle, as did the mean peak plantarflexion moment (Ankle_Moment_X_P). The two peaks were significantly different between conditions (CC peak dorsiflexion angle: 89.0°; SFC peak dorsiflexion: 90.4°; CC peak plantarflexion moment: 1.6 P BWT·m; SFC peak plantarflexion moment: 1.6 P BWT·m). The ankle push-off moment increase may be due to a requirement for an increased magnitude of propulsion in order to swing the weightier lower limb, which is encompassed by the frame. However, the plantarflexion angle showed a change of just 2˚ suggesting that although the difference was significant, it is likely to be of minimal clinical significance. The increased peak knee extensor moment seen for the SFC will have assisted at this propulsive phase of the gait cycle. The reduced step length on the leg with the frame, suggests that although the ankle plantarflexion and knee extension moments increased, there is a lack of propulsive force to swing the limb forward in the same way as without the frame[24]. The increased plantarflexion moment consequently lead to an observable kinematic alteration at the ankle.

The abduction-adduction angle at the knee showed two small peaks, an initial abduction and a secondary adduction peak. Adduction at the knee indicates movement of the distal thigh towards the midline of the body and the distal shank away from the body. Therefore, abduction at the knee represents medial movement of the shank, into a more varus position. The mean peak knee abduction angle (Knee_Angle Y_Ab) increased when wearing the frame, whereas the adduction peak (Knee_Angle Y_Ad) decreased (both showing significant differences between test conditions). Results suggest that the shank is likely to adopt a more varus position when wearing the frame, which may be related to the increased stride width shown for the SFC[25] (CC: 0.11 m; SFC: 0.14 m). Again, the weight of the frame may have influenced this adaptation. Abduction-adduction knee moment data showed an adduction peak and two abduction peaks. Peak knee abduction moments were found to be significantly different between test conditions, whereas the peak knee adduction moments were not (Table 1). Both mean abduction peaks showed a decreased magnitude for the SFC, compared to the CC (Table 1). This decrease in knee abduction moments seen for the SFC is to be expected when considering the adoption of a more varus position of the shank, when wearing the frame. The kinematic changes that occurred when wearing the frame will influence the line of axes, between the hip and the ankle, and therefore influence bending moments and loading.

Biomechanical alterations may influence the bone healing and remodelling seen at the fracture site for patients, as the line of action of the force will be different when walking with and without the Ilizarov frame, although clinically this impact will be minimal. From an engineering perspective, the alterations highlight that there may be scope for the redesign of the frame in order to decrease size and weight.

Hypothesis 1 can be accepted as there was a clear alteration in stride width and lower limb kinematics. Hypothesis 2 must be rejected, as although significant alterations were seen for hip and ankle moments, changes were small and not likely to be clinically significant.

Limitations: No account was taken of the effect of injury, pain or the pins attaching the frame to the bone which will potentially tether soft tissues and lead to pain and joint stiffness. Patients were not included in the study as the overall aim was to isolate the effects of the presence of the frame itself. However, it is important to appreciate that a patient group is likely to show different findings to a healthy group and may even magnify the effects seen in this study, due to the injury itself.

Summary: When summarising the changes seen for subjects when wearing the simulated frame (compared to without), four major points can be identified:

1) Walking velocity was decreased for individuals when wearing the frame, which can be attributed to the decreased step length for the limb with the attached frame.
2) Step width was significantly changed (mean increase) when wearing the frame. This may have led to further kinematic alterations, particularly at the knee, with an increased abduction angle and net adduction moment in the sagittal plane.
3) There was a clear change relating to the push-off phase, when wearing the frame. Significant kinematic changes were identified at this point, with a mean increase in ankle dorsiflexion angle. Additionally a mean increase and significant difference in both ankle plantarflexion and knee extension moments was identified. These findings can be attributed to the increased inertia of the leg with the frame attachment, leading to an increased requirement of force in order to swing the leg forwards.

It is relevant to note that many of the kinematic changes identified in the present study, although some with medium and large effect sizes, in fact showed small differences between the mean data sets. This may explain why differences in peak vertical GRF between the two test conditions were insignificant.

Wearing an Ilizarov frame will lead to small and clinically insignificant changes to an individual's biomechanics, independent to injury. The small adaptations are likely due to the increase in size and weight at the lower limb with the frame. However, if healthy subjects have shown adaptations when wearing the frame it is possible that changes will be magnified when the frame is bolted to an injured, painful leg. This may be particularly relevant when considering patients with considerable muscle damage and pain at the injury site. This, however, is difficult to predict based upon the data presented. The study provides an argument for potentially re-designing the Ilizarov frame in a way that reduces the diameter and mass of the structure. Having said this, it is crucial to keep key elements of the device, such as the tension and positioning of wires, to ensure that the device continues to provide a reliable method for fracture healing. Additionally, the results provide beneficial information for both patients and physiotherapists, when introducing weight bearing activity following the frame attachment. Further research should compare fracture patient population groups and different device designs in order to fully understand the impact of external fixation on gait.

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The Authors declare that there is no conflict of interest

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