UNIVERSITAT DE VIC – UNIVERSITAT CENTRAL DE CATALUNYA

EFECTO DE LA ORTESIS PLANTAR EN LA CINEMÁTICA DEL PIE DURANTE LA CARRERA MEDIANTE SENSORES INERCIALES

Programa de doctorado: Ciencias Experimentales y Tecnología

Juan Luis Florenciano Restoy

Directores de tesis

Dr. Jordi Solé-Casals

Dra. Xantal Borràs-Boix

2023

Dedicado…

A mi familia en general por su apoyo y en particular a mi esposa, por su infinita paciencia durante los años que ha durado esta tesis.

Agradecimientos

La realización de este doctorado ha sido posible por la colaboración inestimable de tres personas. Entre ellas, forman parte destacada la de mis dos directores de Tesis, en primer lugar, la Dra. Xantal Borràs y en segundo lugar el Dr. Jordi Solé, que me han ayudado en todo momento a poder integrar bajo sus conocimientos la sistemática, metodología y aportación científica, tan específica de este doctorado. La Dra. Xantal Borràs por su capacidad y conocimiento de la biomecánica humana, también destacar su capacidad de trabajo para integran en los documentos publicados tanto la forma como el fondo que han servido sin duda para elevar el nivel académico de esta tesis. También dar las gracias al Dr. Jordi Solé por su colaboración para el tratamiento de las señales del sensor inercial y su experiencia en el tema, enfoque y precisión de los trabajos en las diferentes revistas de impacto científico. Para terminar, debo mencionar la ayuda desinteresada y entrañable de mi hijo Juan Luis Florenciano Merino (Ingeniero Aeronáutico y Dr. En Mecánica de Fluidos) muy conocedor del programa de matemáticas MATLAB, con el que se han podido obtener mediante los algoritmos pertinentes datos imprescindibles y resultados muy interesantes desde el punto de vista científico y podológico.

Por último, dar las gracias al órgano responsable de la escuela de doctorado de la UVIC por su apoyo, asesoramiento y compromiso, en los años que ha durado este proyecto.

TABLA DE CONTENIDOS

Índice de figuras y tabla

Figuras

Tablas

Índice de abreviaturas

ABREVIATURA	SIGNIFICADO
2D	Bidimensional
3D	Tridimensional
ABD-ADD	Abducción-aducción
CGC	Centro de gravedad del cuerpo
COP	Centro de presión
D-PF	Flexión dorsi-plantar
EVA	Etileno-acetato de vinilo
EV-INV	Eversión-inversión
F-DP	Flexión dorso-plantar
FLEX-EXT	Flexión-Extensión
FLEX-P	Flexión plantar
FOs	"Foot Orthoses"
FPI-6	Índice de postura del pie
FSP	Patrón de pisada
GP	Patrón de marcha
GRF	Fuerza de reacción del suelo
IMU	Unidad de medida inercial
IMU$_s$	Unidades de medida inercial
MTSS	Síndrome de estrés tibial
OP	Ortesis Plantar
UVIC-UCC	Universistat de Vic - Universitat Central de Catalunya
V2D	Análisis de video bidimensional
V3D	Análisis de video tridimensional
ZUPT	Velocidad del pie 0

Resumen

Observar la cinemática del pie mediante la utilización de sensores inerciales es un campo de estudio de enorme interés para la podología deportiva. Los sensores compuestos por acelerómetros, giroscopios y magnetómetros son capaces de medir la amplitud y velocidad angular del pie en la carrera.

La ortesis plantar (OP) se diseñan específicamente para proporcionar soporte y mejorar la alineación del pie y con ello la alineación de la extremidad. En el atleta, es fundamental entender como las ortesis plantares deportivas a medida pueden influir en la forma en que un corredor se mueve y se desplaza.

Esta tesis doctoral presenta tres artículos previamente publicados sobre la cinemática del pie mediante la utilización de sensores inerciales, que quieren dar respuesta a los objetivos generales de:

1. Analizar la cinemática del pie mediante IMU en corredores aficionados sanos.
2. Determinar si existen asimetrías entre extremidades en las variables cinemáticas con y sin ortesis plantar.
3. Determinar el efecto de las ortesis plantares sobre la cinemática del pie.

Material y Método. Participaron en el estudio un total de 40 varones (43,0 ± 13,8 años) sin lesiones locomotoras en el momento de la prueba. Los sensores se sujetaron con velcro® y cinta adhesiva al dorso del calzado deportivo. Corrieron en una cinta de correr durante 20 segundos a una velocidad de 2,5 m/s. Primero con su calzado habitual y después el mismo tiempo y a la misma velocidad con su calzado habitual más ortesis plantar deportiva.

Estudio 1. Tuvo como objetivo realizar una descripción cinemática del movimiento del pie en el ciclo completo de la carrera en los tres planos del espacio y determinar si hay diferencias entre el pie izquierdo y pie derecho. Los resultados permiten observar que durante la carrera el pie mantiene una secuencia lógica de movimiento, y que existen diferencias con más grados de amplitud de movimiento en pie derecho que en pie izquierdo en los planos frontal y transverso. En el plano sagital no hay diferencias en la amplitud de movimiento.

Estudio 2. Los objetivos de este estudio fueron (1) determinar la amplitud de movimiento del pie en los tres planos de movimiento del pie, solo en la fase de apoyo, (2) Determinar la existencia de asimetrías entre los pies y (3) evaluar los efectos de las ortesis de pie en la cinemática del pie. Se observa una simetría del movimiento del pie en el plano sagital, pero no en los planos frontal y transverso. Las OP no corrigen las asimetrías observadas en estos dos planos de movimiento. Existe una tendencia de la ortesis plantar a reducir los grados de amplitud en los movimientos de los planos frontal y transverso. El tiempo de contacto del apoyo y el tiempo total del ciclo de paso es mayor con el uso de OP, no obstante, la frecuencia de paso no se ve alterada, sugiriendo que las OP no interfieren sobre la técnica de carrera.

Estudio 3. Este estudio tiene como objetivo (1) describir el comportamiento de la velocidad angular del pie en el ciclo de zancada y (2) determinar el efecto de la ortesis plantar en la velocidad angular en los tres planos del movimiento del pie. Los cambios cinemáticos en la velocidad del pie ocurren entre el 30% y el 60% del ciclo. La OP reduce la velocidad angular en el movimiento de abducción y flexión dorsi-plantar, mientras que aumenta la velocidad en el movimiento de inversión-eversión, lo que facilita la transición a la pierna oscilante y con ello el desplazamiento del centro de masa.

Conclusión general. Aunque el patrón de movimiento es similar entre pie derecho y pie izquierdo, los resultados generales permiten observar que existen diferencias, con más grados de amplitud, de movimiento entre el pie derecho en los planos frontal y transverso. La tendencia de la OP es el de reducir los grados de amplitud de movimiento en los planos frontal y transverso, con mayor tiempo de apoyo y ciclo completo con ortesis plantar y una disminución en el número de zancadas, pero no a corregir las posibles asimetrías que pueden existir entre los dos pies. La velocidad angular del pie se ve afectada a partir de la fase estacionaria, por la inercia de la pierna oscilante, entre el 30% y 60% del ciclo. La OP modifica la cinemática disminuyendo la velocidad angular del pie en los planos sagital y transverso, y aumentando la velocidad angular del pie en el plano frontal. Todo ello ayudaría a mejorar la transición hacia la pierna oscilante y con ello tanto el gasto energético como el equilibrio dinámico de la carrera.

Resum

Observar la cinemàtica del peu mitjançant la utilització de sensors inercials és un camp d'estudi d'interès enorme per a la podologia esportiva. Els sensors compostos per acceleròmetres, giroscopis i magnetòmetres són capaços de mesurar l'amplitud i la velocitat angular del peu a la cursa.

L'ortesi plantar (OP) es dissenyen específicament per proporcionar suport i millorar l'alineació del peu i l'alineació de l'extremitat. A l'atleta, és fonamental entendre com les ortesis plantars esportives a mida poden influir en la manera com un corredor es mou i es desplaça.

Aquesta tesi doctoral presenta tres articles prèviament publicats sobre la cinemàtica del peu mitjançant la utilització de sensors inercials, que volen donar resposta als objectius generals de:

1. Analitzar la cinemàtica del peu mitjançant IMU en corredors aficionats sans.
2. Determinar si hi ha asimetries entre extremitats en les variables cinemàtiques sense ortesi plantar.
3. Determinar l'efecte de les ortesis plantars sobre la cinemàtica del peu.

Material i Mètode. Van participar a l'estudi un total de 40 homes (43,0 ± 13,8 anys) sense lesions locomotores en el moment de la prova. Els sensors es van subjectar amb velcro® i cinta adhesiva al dors del calçat esportiu. Van córrer en una cinta de córrer durant 20 segons a una velocitat de 2,5 m/s. Primer amb el calçat habitual i després el mateix temps ia la mateixa velocitat amb el calçat habitual més ortesi plantar esportiva.

Estudi 1. Va tenir com a objectiu fer una descripció cinemàtica del moviment del peu en el cicle complet de la carrera en els tres plans de l'espai i determinar si hi ha diferències entre el peu esquerre i el peu dret. Els resultats permeten observar que durant la cursa el peu manté una seqüència lògica de moviment, i que hi ha diferències amb més graus d'amplitud de moviment en peu dret que en peu esquerre als plans frontal i transvers. Al pla sagital no hi ha diferències en l'amplitud de moviment.

Estudi 2. Els objectius d'aquest estudi van ser (1) determinar l'amplitud de moviment del peu als tres plans de moviment del peu, només a la fase de suport, (2) Determinar

l'existència d'asimetries entre els peus i (3) avaluar els efectes de les ortesis de peu a la cinemàtica del peu. S'observa una simetria del moviment del peu al pla sagital, però no als plans frontal i transvers. Les OP no corregeixen les asimetries observades en aquests dos plans de moviment. Hi ha una tendència de l'ortesi a reduir els graus d'amplitud en els moviments dels plans frontal i transvers. El temps de contacte del suport i el temps total del cicle de pas és més gran amb l'ús d'OP, però la freqüència de pas no es veu alterada, suggerint que les OP no interfereixen sobre la tècnica de carrera.

Estudi 3. Aquest estudi té com a objectiu (1) descriure el comportament de la velocitat angular del peu al cicle de gambada i (2) determinar l'efecte de l'ortesi plantar a la velocitat angular en els tres plans del moviment del peu. Els canvis cinemàtics en la velocitat del peu tenen lloc entre el 30% i el 60% del cicle. L'OP redueix la velocitat angular en el moviment d'abducció i flexió dorsi-plantar, mentre que augmenta la velocitat en el moviment d'inversió-eversió, cosa que facilita la transició a la cama oscil·lant i amb això el desplaçament del centre de massa.

Conclusió general. Tot i que el patró de moviment és similar entre peu dret i peu esquerre, els resultats generals permeten observar que hi ha diferències, amb més graus d'amplitud, de moviment entre el peu dret als plans frontal i transvers. La tendència de l'OP és reduir els graus d'amplitud de moviment en els plans frontal i transvers, amb més temps de suport i cicle complet amb ortesi plantar i una disminució en el nombre de gambades, però no a corregir les possibles asimetries que poden existir entre els dos peus. La velocitat angular del peu es veu afectada a partir de la fase estacionària, per la inèrcia de la cama oscil·lant, entre el 30% i el 60% del cicle. L'OP modifica la cinemàtica disminuint la velocitat angular del peu als plans sagital i transvers, i augmentant la velocitat angular del peu al pla frontal. Tot això ajudaria a millorar la transició cap a la cama oscil·lant i amb això tant la despesa energètica com l'equilibri dinàmic de la cursa.

Abstract

Observing the kinematics of the foot through the use of inertial sensors is a field of study of enormous interest for sports podiatry. Sensors composed of accelerometers, gyroscopes and magnetometers are capable of measuring the amplitude and angular speed of the foot during the run.

Plantar orthoses (OP) are specifically designed to provide support and improve the alignment of the foot and thus the alignment of the limb. For the athlete, it is essential to understand how custom sports plantar orthoses can influence the way a runner moves and moves.

This doctoral thesis presents three previously published articles on the kinematics of the foot through the use of inertial sensors, which aim to respond to the general objectives of:

1. To analyze foot kinematics using IMU in healthy amateur runners.
2. Determine if there are asymmetries between extremities in the kinematic variables with and without plantar orthosis.
3. Determine the effect of plantar orthoses on foot kinematics.

Material and method. A total of 40 men (43.0 ± 13.8 years) without locomotor injuries at the time of testing participated in the study. The sensors were attached with Velcro and adhesive tape to the back of the sports shoe. They ran on a treadmill for 20 seconds at a speed of 2.5 m/s. First with your usual footwear and then the same time and at the same speed with your usual footwear plus sports plantar orthosis.

Study 1. Its objective was to carry out a kinematic description of the movement of the foot in the complete running cycle in the three planes of space and to determine if there are differences between the left foot and the right foot. The results allow us to observe that during running the foot maintains a logical sequence of movement, and that there are differences with more degrees of amplitude of movement in the right foot than in the left foot in the frontal and transverse planes. In the sagittal plane there are no differences in range of motion.

Study 2. The objectives of this study were (1) to determine the range of motion of the foot in the three planes of movement of the foot, only in the stance phase, (2) to determine the existence of asymmetries between the feet and (3) evaluate the effects of foot orthoses on foot kinematics. Symmetry of foot movement is observed in the sagittal plane, but not in the frontal and transverse planes. OPs do not correct the asymmetries observed in these two planes of movement. There is a tendency for the plantar orthosis to reduce the degrees of amplitude in the movements of the frontal and transverse planes. The contact time of the support and the total time of the step cycle is greater with the use of OP; however, the step frequency is not altered, suggesting that the OP does not interfere with the running technique.

Study 3. This study aims to (1) describe the behavior of foot angular velocity in the stride cycle and (2) determine the effect of the plantar orthosis on angular velocity in the three planes of foot movement. Kinematic changes in foot speed occur between 30% and 60% of the cycle. The OP reduces the angular velocity in the abduction and dorsi-plantar flexion movement, while it increases the speed in the inversion-eversion movement, which facilitates the transition to the swing leg and thus the displacement of the center of mass.

General conclusion. Although the movement pattern is similar between the right foot and the left foot, the general results allow us to observe that there are differences, with greater degrees of amplitude, of movement between the right foot in the frontal and transverse planes. The tendency of OP is to reduce the degrees of range of motion in the frontal and transverse planes, with greater support time and complete cycle with plantar orthosis and a decrease in the number of strides, but not to correct the possible asymmetries that They can exist between the two feet. The angular velocity of the foot is affected from the stationary phase, by the inertia of the swinging leg, between 30% and 60% of the cycle. OP modifies the kinematics by decreasing the angular velocity of the foot in the sagittal and transverse planes and increasing the angular velocity of the foot in the frontal plane. All of this would help to improve the transition to the swinging leg and with it both energy expenditure and the dynamic balance of the race.

CAPÍTULO 1

CAPÍTULO 1. TESIS POR COMPENDIO DE TRABAJOS PUBLICADOS

1.1. Estudios publicados.

En la presente tesis doctoral, de acuerdo con el informe correspondiente, autorizado y validado por los directores de tesis y el órgano responsable del programa de doctorado de la Universitat de Vic - Universitat Central de Catalunya (UVic-UCC), se expone como un compendio de tres trabajos previamente publicados. Las referencias completas de los artículos que pertenecen al cuerpo de la tesis son los siguientes:

- **ESTUDIO 1: Florenciano Restoy**, J. L.; Solé-Casals, J.; Borràs-Boix, X. Descriptive analysis of 3D foot motion using inertial sensors: comparison between lower extremities Revespod. 2021;32(1):13-17.
- **ESTUDIO 2: Florenciano Restoy**, J. L.; Solé-Casals, J.; Borràs-Boix, X. IMUs-based effects assessment of the use of foot orthoses in the stance phase during running and asymmetry between extremities. Sensors. 2021. 21, 3277.
- **ESTUDIO 3: Florenciano Restoy**, J. L.; Solé-Casals, J.; Borràs-Boix, X. Effect of foot orthoses on angular velocity of feet. Sensors. 2023. 10, 3390.

1.2. Comunicaciones presentadas.

- **Florenciano Restoy,** J.L. Descriptive analysis of 3D foot motion using inertial sensors: comparison between lower extremities. XXVII Jornadas Científicas de Podología. Abril de 2022.
- **Florenciano Restoy,** J. L.; Solé-Casals, J.; Borràs-Boix, X. Comparative Analysis of Knee Movement with Inertial Sensor. International Workshop on Higher Education. Doctoral Programme in Experimental Sciences and Technology. 5-2017.

1.3. Motivación de este estudio.

La motivación detrás de esta investigación radica en el deseo de comprender cómo el uso de ortesis plantares deportivas a medida puede afectar directamente la cinemática de la carrera de un atleta. La cinemática se refiere al estudio del movimiento en términos de posición, amplitud, velocidad y aceleración.

En el contexto del deporte y el running, es fundamental entender como las ortesis plantares deportivas a medida, pueden influir en la forma en que un corredor se mueve y se desplaza. Estos dispositivos están diseñados específicamente para proporcionar soporte y mejorar la alineación del pie y con ello la alineación de la extremidad, lo que podría tener un impacto significativo en la cinemática del corredor.

Al realizar mediciones cinemáticas utilizando sensores inerciales, es posible obtener datos precisos sobre el movimiento de los corredores antes y después de usar las ortesis plantares deportivas. Esto permite una evaluación objetiva de las posibles modificaciones en la cinemática y su relación con la biomecánica de la carrera.

El objetivo de esta investigación es determinar si el uso de ortesis plantares deportivas a medida produce cambios positivos en la cinemática del corredor. Estas mejoras podrían incluir una mayor eficiencia en la zancada, una distribución de la carga más equilibrada en los pies, una reducción en el riesgo de lesiones y un rendimiento general mejorado.

En resumen, la motivación de esta tesis radica en el interés por utilizar la capacidad de medida cinemática de los sensores inerciales para investigar cómo las ortesis plantares deportivas a medida pueden modificar la cinemática del corredor y, a su vez, mejorar la biomecánica de la carrera. A través de esta investigación, se busca proporcionar una base científica sólida que respalde la implementación y el uso efectivo de estas ortesis, con el objetivo de optimizar el rendimiento y prevenir lesiones en los corredores.

CAPÍTULO 2

CAPÍTULO 2. MARCO TEÓRICO

El estudio de la biomecánica humana, desde la cinemática a la cinética tiene amplias aplicaciones en la salud humana, por ejemplo, en el estudio de la marcha, tanto normal como patológica, para la seguridad de los trabajadores mejorando la ergonomía laboral, y por supuesto en el ámbito del deporte, el gesto deportivo de los atletas y la prevención de lesiones (S. J. Lee & Hidler, 2008; Ojeda et al., 2015).

Históricamente, los estudios de investigación de la cinemática humana en todos estos campos están con frecuencia limitados a entornos de laboratorio. La unidad de medida inercial o sensor inercial (IMU), es una herramienta para extraer datos de amplitud y velocidad de movimiento en entornos que no son de laboratorio, como el lugar de trabajo, el hogar, la clínica, la instalación deportiva etc. En la presente tesis, se ha seleccionado el pie como segmento para el estudio con IMU debido a la importancia que tiene en el deporte en general y la carrera en particular (Bourdon et al., 2017; Li et al., 2014; Montesinos et al., 2018; Rucco et al., 2018).

2.1. La carrera humana, parámetros y terminología

La carrera humana es un proceso de locomoción humano en el que el cuerpo se mueve hacia delante, siendo el peso soportado, alternativamente, por ambas extremidades. Al desplazar el cuerpo sobre la pierna de apoyo, la otra extremidad se balancea y se prepara para el siguiente apoyo (Sánchez Lacuesta, 1999). La diferencia con respecto a la marcha es que en el proceso de transferencia del peso del cuerpo de la pierna retrasada a la pierna adelantada ambos pies se separan del suelo, lo que convierte la carrera humana en una sucesión consecutiva de saltos (Sánchez Lacuesta, 1999).

En la carrera es habitual utilizar la siguiente terminología (Sánchez Lacuesta, 1999):

- **Zancada o ciclo de la carrera**: Es la repetición del contacto de un pie con el suelo hasta el nuevo contacto del mismo pie con el suelo, por ejemplo, contacto del pie derecho hasta el nuevo contacto del pie derecho lo que implica dos saltos.
- **Apoyo:** Es el periodo transcurrido en el que el pie está en contacto con el suelo, se divide en fases, inicio de contacto, apoyo medio e impulso.

- **Oscilación:** Es el periodo de avance de la extremidad, se divide en oscilación inicial, oscilación media y oscilación final.
- **Salto:** Es la distancia recorrida entre el apoyo de ambos pies que implica una fase aérea.

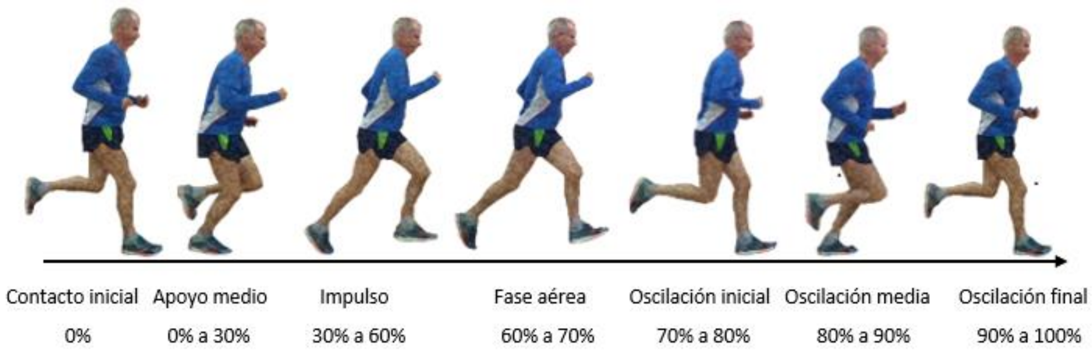


Figura 1. Fases de una zancada de carrera y porcentaje (%) de duración del total del ciclo en carrera a 2.5 m/s. Fuente propia.

En la Figura 1 se observa que al analizar un ciclo completo de carrera el periodo de apoyo se divide en: contacto inicial (0% del ciclo), apoyo medio flexión de la rodilla (30% del ciclo) e impulso extensión de la rodilla (60% del ciclo), seguidamente se contempla la fase aérea (70% del ciclo). La oscilación inicial (80% del ciclo) comienza con la flexión de rodilla, la oscilación media (90% del ciclo) aumenta la flexión de la rodilla y la oscilación final (100% del ciclo) extiende la articulación de la rodilla, en consecuencia, en esta oscilación final culmina con el contacto del pie de nuevo con el suelo y la finalización del ciclo (Hespanhol et al., 2015).

Velocidad de transición de la marcha a la carrera

Bramale & Lieberman, (2004), proponen que, al aumentar la velocidad de desplazamiento durante el movimiento lineal, la carrera es más eficiente que el caminar al utilizar los mecanismos de masa-resorte que intercambia energía cinética y potencial gracias al efecto de tendones y ligamentos, en la flexión de la extremidad, almacena energía de tensión elástica y posteriormente libera la energía en la extensión de la extremidad. La mayoría de los humanos cambian voluntariamente a correr aproximadamente entre los 2,3 y 2,9 m/s (Bramble & Lieberman, 2004).

2.2. La carrera como actividad física

El ejercicio físico tiene beneficios sobre la condición física de la persona, favorece el estado de salud contribuyendo a reducir la obesidad, influye positivamente sobre enfermedades cardiovasculares, así como en patología como la diabetes, y colesterol. (Blair, S. N., et al., 1989).

La carrera es una de las actividades más usadas en los niveles más altos de aptitud física parecen retrasar la mortalidad por todas las causas, principalmente debido a las tasas más bajas de enfermedades cardiovasculares y cáncer. Se ha podido constatar que el correr se asocia significativamente a menor riesgo de mortalidad por causas de origen, cardiovascular y cáncer, comparado con grupos que no corren. Cualquier cantidad de carrera, incluso una vez por semana, es mejor que no correr (Pedisic et al., 2020).

Hespanhol et al. (2015) llegaron a la conclusión mediante un metaanálisis que la carrera de resistencia fue efectiva al proporcionar efectos beneficiosos sobre la masa corporal, la grasa corporal, la frecuencia cardíaca en reposo, el VO_2 máximo, los triglicéridos y el colesterol HDL en adultos físicamente inactivos. Cuanto mayor es la duración del entrenamiento aeróbico, mayores son los beneficios para la salud y sugieren que tanto los médicos, como las autoridades sanitarias, pueden utilizar esta información para aconsejar a las personas sobre los beneficios sobre la salud.

2.3. Lesiones deportivas por sobrecarga mecánica en la carrera

Sin perder de vista los beneficios para la salud que supone el caminar y correr, existe preocupación en la medicina deportiva por incremento de las lesiones en el sistema locomotor, principalmente cadera, rodilla y pie (Dugan & Bhat, 2005).

Jauhiainen et al. (2020) evaluaron los datos cinemáticos tridimensionales de las articulaciones de la extremidad inferior a 291 corredores con lesión y sin lesión, los resultados sugieren, en contra de la hipótesis de los autores, que la ubicación de la lesión no está relacionada con patrones cinemáticos por el mal funcionamiento de la extremidad inferior, sin embargo, observaron patrones de carrera comunes en todos los corredores independiente de si tenían lesión o no la tenían.

Por lo general el corredor aficionado compite en esfuerzos esporádicos de alta intensidad lo que le predispone a un mayor riesgo de lesiones (Altman & Davis, 2012), entrena una media de 35-40 km/semana lo que implica aproximadamente dos millones de impactos al año. En cada impacto este corredor debe disipar una fuerza aproximada dos veces y medio su peso corporal (Cavanagh & Lafortune, 1980).

Al contemplar estos datos, el corredor de larga distancia está expuesto a lesiones por sobrecarga de las extremidades inferiores con tasas de lesiones que van desde 19,4 % al 79,3 % (Malisoux et al., 2015; Van Gent et al., 2007).

La acumulación de impactos repetitivos en los apoyos por debajo del umbral de lesión aguda produce un efecto de fatiga que mantenida en el tiempo forma parte de las lesiones por sobrecarga mecánica (Glauberman & Cavanagh, 2014).

Otra cuestión como factor de riesgo importante son los corredores que reanudan su actividad después de una lesión, están en riesgo más elevado y deben prestar atención a medidas preventivas (Malisoux et al., 2015).

Se ha podido asociar con el desarrollo de microfracturas en la tibia con los altos índices de carga (Glauberman & Cavanagh, 2014; Schaffler et al., 1989). Varias investigaciones postulan aumento de picos verticales de impacto y altas tasas de carga en corredores que habían padecido fractura por estrés de la tibia (Milner et al., 2006). En una revisión sistemática, Zadpoor & Nikooyan (2011) apoyaron los resultados anteriores al afirmar que las altas cargas verticales se relacionaban con fracturas por estrés en extremidades inferiores. Sin embargo, otros estudios refieren que la zona más afectada por lesiones frecuentes en la carrera, se localizan en la rodilla, le sigue la pierna, el pie y la zona del muslo, el dolor patelo-femoral, síndrome de la banda iliotibial, periostitis tibial, fascitis plantar y tendinitis del Tendón de Aquiles son las más referidas (Van Gent et al., 2007).

En un ensayo sobre el síndrome de estrés tibial (MTSS) los investigadores hallaron que la eversión y abducción del antepié y el retropié diferían entre sujetos con y sin MTSS al correr (Okunuki et al., 2019). Además, los sujetos con MTSS tuvieron mayor abducción del antepié y eversión y abducción del retropié durante la ejecución que aquellos sin MTSS. Por lo que concluyeron que para la prevención y rehabilitación de MTSS, puede ser importante dar un enfoque, no solo en la parte posterior del pie sino también, en la cinemática del antepié al correr (Okunuki et al., 2019).

Actualmente las investigaciones no aclaran los factores predictivos de lesión en los corredores populares, lo cual limita el conocimiento de los factores de riesgo y las estrategias para la prevención (Buist et al., 2010).

Nigg et al., (2017) propone un cambio de paradigma, según sus trabajos de investigación no hay evidencia epidemiológica o funcional concluyente que demuestre que la pronación en si misma sea la causa de lesión, al ser un movimiento necesario y que, por lo tanto, el paradigma de la pronación como causa de lesión es cuestionable.

La evidencia prospectiva actual que relaciona las variables biomecánicas con el riesgo de lesiones es escasa e inconsistente, y los hallazgos dependen en gran medida de la población y las lesiones estudiadas. Se necesita investigación futura para confirmar estos factores de riesgo biomecánicos y determinar si la modificación de estas variables puede ayudar en la prevención y el manejo de lesiones (Ceyssens et al., 2019).

2.4. La ortesis plantar en las lesiones deportivas

Se ha relacionado el efecto de la ortesis de pie a la reducción de la eversión del retropié, de modo que esto pueda asociarse con una reducción del momento de aducción de la rodilla durante la primera fase de apoyo, mejorando la estabilidad de la extremidad (Eslami et al., 2009).

Papuga & Burke, (2011) proponen en un estudio que el uso de OP mejoran las condiciones de apoyo en atletas y están siendo eficaces en el tratamiento de las lesiones por sobrecarga mecánica vinculados con las variaciones estructurales del pie y sugieren que el mayor efecto de la ortesis deportiva es en la fase de apoyo medio, aproximadamente el 50% del apoyo del pie, condicionado por el movimiento axial de la tibia.

La ortesis plantar deportiva a medida es un dispositivo externo que se utiliza en la planta del pie junto al calzado deportivo aumentando así el nivel de comodidad y amortiguación, por lo que deberían estar diseñadas específicamente con la suficiente comodidad para mejorar la ruta preferida por el sujeto en su sistema nervioso durante la carrera (Nigg et al., 1999). Nigg et al. (2017) proponen que el movimiento de pronación es un movimiento natural del tobillo donde el pie se mueve en flexión eversión y abducción y por lo tanto la pronación no debe ser siempre la causa de la lesión.

Mündermann et al., (2003) comprobaron que la colocación de OP redujo la eversión máxima del pie y el momento de inversión del tobillo, modificando la carga vertical y momento máximo de rotación interna de la rodilla. Lack et al., (2014), mediante la colocación de ortesis anti-pronación del pie, utilizadas durante la prueba, redujeron la aducción de cadera y la rotación interna de la rodilla después del contacto inicial del pie en individuos con pronación severa.

Franklyn-Miller et al. (2011) evaluaron corriendo a cuatrocientos aspirantes a oficiales, mediante registro en placa de presión de las presiones de sus pies. De las dos variables los que utilizaron OP personalizadas redujeron la frecuencia de lesiones.

2.5. Asimetría en la carrera

Bache & Orellana (2014) definen la lateralidad como la preferencia sistemática de utilización de uno u otro órgano par del cuerpo (ojos, oídos, manos, pies…) en las actividades de la vida diaria y se trata de un fenómeno existente tanto en seres humanos como en animales

Vagenas & Hoshizaki (1991) demostraron en un estudio que relacionaba la lateralidad con las asimetrías anatómicas del pie (rango de movimiento de la articulación subtalar: inversión, eversión, flexibilidad pasiva talo-calcánea y rango total de movimiento), que existían asimetrías de fuerza (pico isocinético en la flexo-extensión de rodilla) entre la pierna dominante y la contralateral en corredores de larga distancia. Estas asimetrías se atribuían, factores biomecánicos y fisiológicos (diferencias en la adaptación neuromuscular, activación de unidades motoras y distribución de las fibras rápidas y lentas), por lo que la dominancia podría entenderse como una particularidad muy asociada a la tarea en lugar de un fenómeno general para una extremidad completa.

Gao et al. (2022) demostraron que la presión medial del pie del miembro dominante fue mayor que la del miembro no dominante. El rango del índice de equilibrio del pie mostró asimetría en las tareas de caminar y correr (índice de simetría absoluta <10%). No hubo diferencia significativa en la masa muscular entre los miembros inferiores dominantes y no dominantes (p = 0,79). La cuantificación de las posibles diferencias y asimetrías podría proporcionar implicaciones para la prevención de lesiones en la marcha y el diseño del calzado.

En la carrera de larga distancia es común que las extremidades inferiores no impactan en el suelo con la misma intensidad. Esto se debe a que generalmente se puede distinguir un miembro inferior que actúa como amortiguador y otro que actúa como propulsor (Viel, 2002).

Sadeghi et al. (2000) observaron diferencias funcionales entre ambas extremidades probablemente relacionadas con la función de cada miembro en tareas de propulsión y control durante la marcha sin discapacidad. En los debates actuales sobre la simetría de la marcha en personas sin discapacidad, se ha citado la lateralidad como explicación de la existencia de diferencias funcionales entre las extremidades inferiores, aunque varios estudios no apoyan la hipótesis de una relación entre la simetría de la marcha y la lateralidad. Se necesita más investigación para demostrar la asimetría funcional de la marcha y su relación con la lateralidad, teniendo en cuenta aspectos biomecánicos de la marcha.

Polk et al. (2017) comprobaron en sus resultados que los componentes GRF verticales, de frenado y de propulsión eran gran medida simétricos, pero existían asimetrías significativas en las fuerzas e impulsos máximos mediolaterales con componentes de GRF dirigidos lateralmente más altos y medialmente más bajos generados por las extremidades derechas dominantes.

2.6. Análisis de la cinemática del pie mediante IMU

La cinemática del pie se refiere al estudio de los movimientos y posiciones del pie durante un ciclo completo de carrera u otras actividades físicas. Se centra en analizar los diferentes componentes biomecánicos del pie, así como, las articulaciones implicadas, por un lado, las de soporte y amortiguación: Subastragalina, Chopart, Lisfranc y, por otro lado, las articulaciones de movimiento que son: la del tobillo y las metatarsofalángicas (Viladot Pericé, 2000).

Para procesar los datos del sensor inercial la colocación del IMU es importante. Autores como Boutaayamou et al. (2015) utilizaron dos dispositivo IMU para el estudio del pie, uno colocado en el talón de la zapatilla deportiva y el otro en la parte proximal del dedo gordo, con ello determinaron parámetros cinemáticos de la extremidad.

Bötzel et al., (2016) pudieron determinar que el contacto inicial del pie provocó alteración en ambos acelerómetros de cada extremidad. El algoritmo de Matlab identificó la señal del giroscopio de manera confiable después de que el ruido de estos datos hubiera sido eliminado por un filtro de retardo de fase cero de paso bajo de 20 Hz.

Trojaniello et al. (2014) presentaron un método, utilizando IMU, donde la velocidad de la marcha no afectó sustancialmente su rendimiento. Las estimaciones de los parámetros espaciotemporales mostraron errores más pequeños que los informados en estudios anteriores y un nivel similar de precisión y exactitud para los patrones de marcha tanto sanos como patológicos. La combinación de robustez, precisión y exactitud sugiere que el método propuesto es adecuado para el uso clínico habitual.

2.7. Bibliografía

Altman, A. R., & Davis, I. S. (2012). Barefoot running: Biomechanics and implications for running injuries. Current sports medicine reports, 11(5), 244-250.

Bache, M. A. B., & Orellana, J. N. (2014). Lateralidad y rendimiento deportivo. Archivos de medicina del deporte: revista de la Federación Española de Medicina del Deporte y de la Confederación Iberoamericana de Medicina del Deporte, 31(161), 200-204.

Bötzel, K., Marti, F. M., Rodríguez, M. Á. C., Plate, A., & Vicente, A. O. (2016). Gait recording with inertial sensors–How to determine initial and terminal contact. Journal of biomechanics, 49(3), 332-337.

Bourdon, P. C., Cardinale, M., Murray, A., Gastin, P., Kellmann, M., Varley, M. C., Gabbett, T. J., Coutts, A. J., Burgess, D. J., & Gregson, W. (2017). Monitoring athlete training loads: Consensus statement. International journal of sports physiology and performance, 12(s2), S2-161.

Boutaayamou, M., Schwartz, C., Stamatakis, J., Denoël, V., Maquet, D., Forthomme, B., Croisier, J.-L., Macq, B., Verly, J. G., & Garraux, G. (2015). Development and validation of an accelerometer-based method for quantifying gait events. Medical engineering & physics, 37(2), 226-232.

Bramble, D. M., & Lieberman, D. E. (2004). Endurance running and the evolution of Homo. nature, 432(7015), 345-352.

Buist, I., Bredeweg, S. W., Lemmink, K. A. P. M., Van Mechelen, W., & Diercks, R. L. (2010). Predictors of Running-Related Injuries in Novice Runners Enrolled in a Systematic Training Program: A Prospective Cohort Study. The American Journal of Sports Medicine, 38(2), 273-280. https://doi.org/10.1177/0363546509347985

Cavanagh, P. R., & Lafortune, M. A. (1980). Ground reaction forces in distance running. Journal of biomechanics, 13(5), 397-406.

Ceyssens, L., Vanelderen, R., Barton, C., Malliaras, P., & Dingenen, B. (2019). Biomechanical risk factors associated with running-related injuries: A systematic review. Sports medicine, 49, 1095-1115.

Dugan, S. A., & Bhat, K. P. (2005). Biomechanics and analysis of running gait. Physical Medicine and Rehabilitation Clinics, 16(3), 603-621.

Eslami, M., Begon, M., Hinse, S., Sadeghi, H., Popov, P., & Allard, P. (2009). Effect of foot orthoses on magnitude and timing of rearfoot and tibial motions, ground reaction force and knee moment during running. Journal of Science and Medicine in Sport, 12(6), 679-684.

Franklyn-Miller, A., Wilson, C., Bilzon, J., & McCrory, P. (2011). Foot Orthoses in the Prevention of Injury in Initial Military Training: A Randomized Controlled Trial. The American Journal of Sports Medicine, 39(1), 30-37. https://doi.org/10.1177/0363546510382852

Gao, Z., Mei, Q., Xiang, L., Baker, J. S., Fernandez, J., & Gu, Y. (2022). Effects of limb dominance on the symmetrical distribution of plantar loading during walking and running. Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology, 236(1), 17-23. https://doi.org/10.1177/1754337120960962

Blair, S. N., Harold W. Kohl, Ralph S. Paffenbarger, Debra G. Clark, Kenneth H. Cooper, and Larry W. Gibbons. "Physical fitness and all-cause mortality: a prospective study of healthy men and women." Jama 262, no. 17 (1989): 2395-2401.

Glauberman, M. D., & Cavanagh, P. R. (2014). Rearfoot strikers have smaller resultant tibial accelerations at foot contact than non-rearfoot strikers. Journal of Foot and Ankle Research, 7, 1.

Hespanhol, L. C., Pillay, J. D., Van Mechelen, W., & Verhagen, E. (2015). Meta-Analyses of the Effects of Habitual Running on Indices of Health in Physically Inactive Adults. Sports Medicine, 45(10), 1455-1468. https://doi.org/10.1007/s40279-015-0359-y

Jauhiainen, S., Pohl, A. J., Äyrämö, S., Kauppi, J., & Ferber, R. (2020). A hierarchical cluster analysis to determine whether injured runners exhibit similar kinematic gait patterns. Scandinavian Journal of Medicine & Science in Sports, 30(4), 732-740. https://doi.org/10.1111/sms.13624

Lack, S., Barton, C., Woledge, R., Laupheimer, M., & Morrissey, D. (2014). The immediate effects of foot orthoses on hip and knee kinematics and muscle activity during a functional step-up task in individuals with patellofemoral pain. Clinical Biomechanics, 29(9), 1056-1062.

Lacuesta, J. S. (1999). Biomecánica de la marcha humana normal y patológica. Instituto de Biomecánica de Valencia.

Lee, S. J., & Hidler, J. (2008). Biomechanics of overground vs. Treadmill walking in healthy individuals. Journal of Applied Physiology, 104(3), 747-755. https://doi.org/10.1152/japplphysiol.01380.2006

Li, P., Meziane, R., Otis, M. J.-D., Ezzaidi, H., & Cardou, P. (2014). A Smart Safety Helmet using IMU and EEG sensors for worker fatigue detection. 2014 IEEE International Symposium on Robotic and Sensors Environments (ROSE) Proceedings, 55-60. https://ieeexplore.ieee.org/abstract/document/6952983/

Malisoux, L., Ramesh, J., Mann, R., Seil, R., Urhausen, A., & Theisen, D. (2015). Can parallel use of different running shoes decrease running-related injury risk? Scandinavian Journal of Medicine & Science in Sports, 25(1), 110-115. https://doi.org/10.1111/sms.12154

Milner, C. E., Ferber, R., Pollard, C. D., Hamill, J., & Davis, I. S. (2006). Biomechanical factors associated with tibial stress fracture in female runners. Medicine & Science in Sports & Exercise, 38(2), 323-328.

Montesinos, L., Castaldo, R., & Pecchia, L. (2018). Wearable inertial sensors for fall risk assessment and prediction in older adults: A systematic review and meta-analysis. IEEE transactions on neural systems and rehabilitation engineering, 26(3), 573-582.

Mündermann, A., Nigg, B. M., Humble, R. N., & Stefanyshyn, D. J. (2003). Foot orthotics affect lower extremity kinematics and kinetics during running. Clinical biomechanics, 18(3), 254-262.

Nigg, B. M., Mohr, M., & Nigg, S. R. (2017). Muscle tuning and preferred movement path–a paradigm shift. Current Issues in Sport Science (CISS), 2, 007-007.

Nigg, B. M., Nurse, M. A., & Stefanyshyn, D. J. (1999). Shoe inserts and orthotics for sport and physical activities. Medicine and science in sports and exercise, 31, S421-S428.

Ojeda, L. V., Rebula, J. R., Kuo, A. D., & Adamczyk, P. G. (2015). Influence of contextual task constraints on preferred stride parameters and their variabilities during human walking. Medical Engineering & Physics, 37(10), 929-936. https://doi.org/10.1016/j.medengphy.2015.06.010

Okunuki, T., Koshino, Y., Yamanaka, M., Tsutsumi, K., Igarashi, M., Samukawa, M., Saitoh, H., & Tohyama, H. (2019). Forefoot and hindfoot kinematics in subjects with medial tibial stress syndrome during walking and running. Journal of Orthopaedic Research, 37(4), 927-932. https://doi.org/10.1002/jor.24223

Papuga, M. O., & Burke, J. R. (2011). Effects of foot orthotics on running economy: Methodological considerations. J Manipulative Physiol Ther, 34(2), 114-118.

Pedisic, Z., Shrestha, N., Kovalchik, S., Stamatakis, E., Liangruenrom, N., Grgic, J., Titze, S., Biddle, S. J., Bauman, A. E., & Oja, P. (2020). Is running associated with a lower risk of all-cause, cardiovascular and cancer mortality, and is the more the better? A systematic review and meta-analysis. British journal of sports medicine, 54(15), 898-905.

Polk, J. D., Stumpf, R. M., & Rosengren, K. S. (2017). Limb dominance, foot orientation and functional asymmetry during walking gait. Gait & posture, 52, 140-146.

Rucco, R., Sorriso, A., Liparoti, M., Ferraioli, G., Sorrentino, P., Ambrosanio, M., & Baselice, F. (2018). Type and location of wearable sensors for monitoring falls during static and dynamic tasks in healthy elderly: A review. Sensors, 18(5), 1613.

Sadeghi, H., Allard, P., Prince, F., & Labelle, H. (2000). Symmetry and limb dominance in able-bodied gait: A review. Gait & posture, 12(1), 34-45.

Schaffler, M. B., Radin, E. L., & Burr, D. B. (1989). Mechanical and morphological effects of strain rate on fatigue of compact bone. Bone, 10(3), 207-214.

Trojaniello, D., Cereatti, A., Pelosin, E., Avanzino, L., Mirelman, A., Hausdorff, J. M., & Della Croce, U. (2014). Estimation of step-by-step spatio-temporal parameters of normal and impaired gait using shank-mounted magneto-inertial sensors: Application to elderly, hemiparetic, parkinsonian and choreic gait. Journal of NeuroEngineering and Rehabilitation, 11(1), 152. https://doi.org/10.1186/1743-0003-11-152

Vagenas, G., & Hoshizaki, B. (1991). Functional asymmetries and lateral dominance in the lower limbs of distance runners. Journal of Applied Biomechanics, 7(4), 311-329.

Van Gent, R. N., Siem, D., van Middelkoop, M., Van Os, A. G., Bierma-Zeinstra, S. M. A., & Koes, B. W. (2007). Incidence and determinants of lower extremity running injuries in long distance runners: A systematic review. British journal of sports medicine, 41(8), 469-480.

Viel, E. (2002). La marcha Humana, la Carrera y el Salto. Biomecánica, exploraciones, normas y alteraciones. Editorial Masson. Barcelona, 2.

Viladot Pericé, A. (2000). Quince lecciones sobre patología del pie. En Quince lecciones sobre patología del pie (pp. 269-269). https://pesquisa.bvsalud.org/portal/resource/pt/sus-9849

Zadpoor, A. A., & Nikooyan, A. A. (2011). The relationship between lower-extremity stress fractures and the ground reaction force: A systematic review. Clinical biomechanics, 26(1), 23-28.

CAPÍTULO 3

CAPÍTULO 3. OBJETIVOS E HIPÓTESIS.

La problemática actual se centra en el conocimiento limitado sobre la cinemática del pie con sensores inerciales y el efecto de la ortesis plantar en corredores aficionados sanos. Aunque existen estudios sobre este tema, aún se desconoce cómo se comporta el pie durante la carrera en individuos sin lesiones o patologías específicas, si bien es cierto que se han realizado investigaciones sobre el uso de ortesis plantares en corredores con lesiones o condiciones específicas, también es cierto que no hay suficientes estudios que aborden como estas ortesis modifican la cinemática del pie en individuos sanos y dentro de la normalidad.

3.1. Objetivos generales

1. Analizar la cinemática del pie mediante IMU en corredores aficionados sanos.
2. Determinar si existen asimetrías entre extremidades en las variables cinemáticas con y sin ortesis plantar.
3. Determinar el efecto de las ortesis plantares sobre la cinemática del pie.

3.2 Objetivos e hipótesis de cada uno de los estudios

ESTUDIO-I.

Objetivo 1. Realizar una descripción cinemática del desplazamiento angular del pie en los tres planos del espacio durante la carrera.

Objetivo 2. Determinar si existen diferencias en el desplazamiento angular entre ambas extremidades.

Hipótesis 2. La hipótesis nula establece que no habrá diferencias en el desplazamiento angular entre extremidades.

ESTUDIO-II.

Objetivo 1. Evaluar la amplitud de desplazamiento angular del pie de ambas extremidades en 3D durante el periodo de apoyo en la carrera.

Hipótesis 1. La hipótesis nula establece que no habrá diferencias entre los pies en el desplazamiento angular del pie en 3D en el periodo de apoyo.

Objetivo 2. Evaluar el efecto de la OP sobre el desplazamiento angular del pie en 3D en el periodo de apoyo durante la carrera.

Hipótesis 2. La hipótesis nula establece que no existirán diferencias en el desplazamiento angular del pie a favor de los OP.

ESTUDIO-III.

Objetivo 1. Describir la velocidad angular del pie en carrera de individuos sanos en la condición no-OP.

Objetivo 2. Determinar si existen diferencias en la velocidad angular entre pie derecho e izquierdo con y sin OP.

Hipótesis 2. La hipótesis nula establece que no habrá diferencias en la velocidad angular entre pie derecho e izquierdo con y sin OP.

Objetivo 3. Determinar el efecto de la OP en la velocidad angular del pie en cada uno de los ejes de movimiento del pie durante la carrera.

Hipótesis 3. La hipótesis nula establece que no habrá diferencias en la velocidad angular del pie cuando se utilice OP.

CAPÍTULO 4

CAPÍTULO 4. MATERIAL Y MÉTODO

4.1. Generalidades de la metodología

Participaron en el estudio un total de 40 varones (43,0 ± 13,8 años) sin lesiones locomotoras en el momento de la prueba. Las características físicas de los sujetos se registraron mediante una escala de columna mecánica con un medidor de altura (ADE, Hamburgo, Alemania) (175,5 ± 7,0 cm; 72,0 ± 5,5 kg). Todos ellos tenían experiencia y kilometraje acumulado y ya estaban familiarizados con las OP desde hacía al menos un año. Se les informó de las condiciones del estudio y se firmó un consentimiento informado previo a su participación. Este estudio fue aprobado el Comité de Ética de la Universidad de Vic (UVic-UCC, 09/2016) y siguió la Declaración de Helsinki.

El procedimiento de análisis se inició con la colocación de los sensores en el empeine de la zapatilla deportiva, en cada pie. Esta es una IMU comúnmente utilizada en podología y fisioterapia con un acelerómetro triaxial, giroscopio y magnetómetro (MotionPod 30 Hz, Grenoble, Francia). Además del sistema de cierre de velcro® suministrado, se utilizó cinta adhesiva para fijar mejor el dispositivo y reducir las vibraciones. Los participantes llevaban sus zapatillas habituales para correr, ya que se ha observado que los cambios en la dureza de la zapatilla pueden afectar a la cinemática de las extremidades inferiores (Fischer et al., 2017; Lussiana et al., 2013).

La configuración predeterminada de fábrica se mantuvo en este estudio y, por lo tanto, la frecuencia de muestreo fue de 30 Hz, la velocidad de carrera fue de 2,5 m/s. Los datos sin procesar se filtraron utilizando la configuración predeterminada de fábrica (BioVal—RM Ingenierie, 2008; MPOD1, 2008).

Figura 2. Sensor Motionpod utilizado en el estudio

Los sujetos realizaron una carrera de calentamiento de 3 minutos en una cinta de correr (BH Fitness G6414V SPORT, Álava, España) a 2,5 m/s (9 km/h) sin ninguna inclinación (medida con un nivel de burbuja). El calentamiento también permitió a los corredores familiarizarse con la velocidad de la cinta y el entorno. Correr en cinta es representativo de correr en el suelo (Fellin et al., 2010; Firminger et al., 2018)

Una vez finalizado el periodo de calentamiento, el participante descansó durante dos minutos mientras se explicaba el resto del procedimiento y se calibraba el sensor.

Las IMU se calibraron de acuerdo con las instrucciones del fabricante (MPOD1, 2008). En primer lugar, en el soporte de carga al inicio de la sesión y, tras ser colocado en el empeine, con el participante parado y erguido durante 3 s. Por lo tanto, en un sistema de referencia relativo ortogonal triple, el eje vertical captura los movimientos de abducción-aducción (ABD-ADD) del pie, el eje longitudinal captura los movimientos de inversión-eversión (EV-INV) y los movimientos de flexión dorso-plantar (F-DP) se capturan en el eje medial-lateral.

Se realizaron dos mediciones de 20 s, la primera con los participantes con sus zapatillas de correr habituales y la segunda con sus zapatillas de correr habituales y OP en ambos pies, a una velocidad de 2,5 m/s para cumplir con los requisitos de frecuencia de muestreo de la IMU (MPOD1, 2008). Los OP utilizados para cada participante fueron aquellos que usaban regularmente y con los que ya estaban familiarizados. Siguiendo las recomendaciones del fabricante de la IMU (MPOD1, 2008), la adquisición de datos se realizó durante 20 s después de que la velocidad de la cinta se estabilizara. Este tiempo para garantizar la fiabilidad de los datos y evitar errores de medición causados por la integración de datos a lo largo del tiempo.

El sistema de fabricación de la OP consiste en un molde de yeso húmedo de fraguado rápido, adaptado a través de un sistema de colchón de látex asistido por un sistema neumático. A continuación, se realiza un patrón de papel con la anchura y la longitud del pie que se ajusta a una hoja de polipropileno (3 mm) y se calienta en un horno a 180°. Esto permite que el polipropileno se deforme y luego se ajuste al molde de yeso. El mismo proceso se realiza con etileno-acetato de vinilo (EVA) (6 mm de espesor y dureza 30 Shore). El EVA se calienta hasta 80º. Las formas de EVA y propileno se unen con pegamento aprobado.

4.2. Origen de los datos de la tesis

Bioval (BioVal—RM Ingenierie, 2008) es un programa de captura de movimiento tridimensional que comercializa la empresa RM Ingenierie (Rodez, Francia) con la interfaz inalámbrica (2.4 GHz con rango de transmisión de hasta 30 m y 8 horas de uso). Bioval fue concebido para proporcionar estudio cinemático del movimiento de un segmento, o de varios segmentos del cuerpo humano en tiempo real y sin cable de conexión que utiliza los sensores inerciales denominados MotionPod detallados anteriormente.

Las funciones principales de este software son las de comparar series de ejercicios de rehabilitación con ejercicios previos y estimar la amplitud de movimiento en grados. Es utilizado con frecuencia para aplicaciones biomecánicas, tales como la marcha y la carrera como uso podológico específico en consulta.

Procedimiento de obtención de datos.

Siguiendo el procedimiento establecido por Bioval en su protocolo, los datos obtenidos se procesaron de la siguiente manera. El primer registro fue con el calzado deportivo y el segundo registro con calzado deportivo más OP (Tabla 1), los datos quedaron registrados en un fichero CSV ("Comma Separated Values") en Excel. Bioval permite diferenciar entre pie izquierdo y pie derecho. Se guardaron los datos de cada participante en subapartados uno del pie izquierdo sin y otro subapartado con OP y también del pie derecho sin y con OP.

En el Estudio-I se utilizaron los datos aportados directamente por Bioval de amplitud de movimiento en ciclo completo.

En el Estudio-II y Estudio-III los datos se trasladaron y procesaron con Matlab para determinar la amplitud en apoyo del pie y la velocidad angular del pie.

Tiempo	FLEX-D ↑ FLEX-P ↓	ADD ↑ ABD ↓	EV ↑ INV ↓
0	-0.009087433	-0.008272325	-0.01314841
0.033	-0.001614698	-0.000608581	0.02391136
0.067	0.04629232	-0.0022.3984	-0.03470469
0.1	0.05340667	-0.01004707	-0.000954492
0.134	0.1009222	0.004585813	-0.002812494
0.168	0.05340667	0.005422238	0.008236715
0.201	0.07015056	0.007739278	-0.0002929785
0.234	0.06457957	0.00892782	0.0222801
0.268	0.08180248	0.01047812	0.007261595
0.301	0.1164572	0.02224988	0.008682665
0.335	0.1115287	0.01622182	-0.01798024
0.368	0.1264461	-0.000877265	-0.006796319
0.401	0.1014713	-0.01202144	-0.01368312
0.434	0.1336534	-0.02435783	-0.02243421
0.467	0.1613157	0.0128211	0.01368312
0.5	0.1421782	0.06131098	0.009084248
0.534	0.09493089	0.07470985	0.02798213
0.567	0.007729181	0.04971901	0.06381491
0.601	-0.08007326	-0.01762662	0.2311772
0.634	-0.2751299	-0.5049304	0.658507
0.668	-0.9496339	-3.924.668	0.6362455
0.701	-4.737.855	-1.087.685	2.158.647
0.735	-4.542843	-1.431.922	2.463.474

Tabla 1. Ejemplo de datos obtenidos en Excel aportados por el sensor en archivos .CSV.

Caso real con el número 1700 del pie derecho con calzado. La primera columna están los datos del tiempo. En la segunda columna muestra los datos de FLEX-D y FLEX-P, la columna tercera muestra los datos de ADD-ABD y la columna cuarta los datos EV-INV. La flecha ↑ indica en la curva de amplitud, trayectoria ascendente. La flecha ↓ indica en la curva de amplitud, trayectoria descendente.

Obtención de los grados de amplitud en 3D en el periodo de apoyo.

Los datos se procesaron utilizando Matlab (Mathworks, Natick, MA, EE. UU.). Para cada uno de los participantes el proceso fue el siguiente, tomando como referencia el movimiento en el plano sagital F-DP, se determinó la fase de apoyo para cada paso, que se encontraba entre los puntos de máxima y mínima flexión plantar que es el periodo de apoyo del pie. Así, también se obtienen los puntos inicial y final de la fase de apoyo de cada paso en los planos frontal y transverso. Posteriormente se realizó un ajuste matemático para asegurar que en el F–DP la zona de meseta (apoyo medio) se correspondiera con los 0º del movimiento, y las posibles desviaciones del sensor en los movimientos de EV–INV y ABD- ADD, también ajustaron al inicio en 0º. Esta amplitud angular media de cada participante se obtuvo sumando los movimientos durante el periodo de apoyo de cada zancada realizado dentro del período de recogida de datos de 20 s y dividiéndolos por el número total de pasos y se pudo obtener las medias de amplitud angular de los 40 participantes.

Obtención de la velocidad angular en 3D.

Los datos se procesaron con Matlab (Mathworks, Natick, MA, EE. UU.). Para cada uno de los participantes la velocidad angular en cada eje de F–DP, EV–INV y ABD- ADD se determinó calculando la amplitud media de cada movimiento a lo largo del tiempo. Esta amplitud media de velocidad angular se obtuvo sumando los movimientos durante la fase de apoyo de cada paso realizado dentro del período de recogida de datos de 20 s y dividiéndolos por el número total de pasos, después se obtuvo la media de velocidad angular de los 40 participantes.

4.3. Bibliografía

BioVal—RM Ingenierie. (2008). Manual de Instrucciones. BioVal—RM Ingenierie Movea Motion Copyright ©. Available online: https://www. rmingenierie.net/bioval/ (accessed on 10 May 2021).

Fellin, R. E., Manal, K., & Davis, I. S. (2010). Comparison of lower extremity kinematic curves during overground and treadmill running. Journal of applied biomechanics, 26(4), 407-414.

Firminger, C. R., Vernillo, G., Savoldelli, A., Stefanyshyn, D. J., Millet, G. Y., & Edwards, W. B. (2018). Joint kinematics and ground reaction forces in overground versus treadmill graded running. Gait & posture, 63, 109-113.

Fischer, K. M., Willwacher, S., Hamill, J., & Brüggemann, G.-P. (2017). Tibial rotation in running: Does rearfoot adduction matter? Gait & posture, 51, 188-193.

Lussiana, T., Fabre, N., Hébert-Losier, K., & Mourot, L. (2013). Effect of slope and footwear on running economy and kinematics. Scandinavian Journal of Medicine & Science in Sports, 23(4). https://doi.org/10.1111/sms.12057.

MPOD1. (2008). MotionPod Manual de Usuario. Movea Motion Copyright ©. Available online: https://fccid.io/JJ4-MPOD1 /Users-Manual/Users-Manual-1206708 (accessed on 10 May 2021).

CAPÍTULO 5

CAPÍTULO 5. PRESENTACIÓN DE LOS ARTICULOS PUBLICADOS

5.1. ESTUDIO 1: Descriptive analysis of 3d foot motion using inertial sensors: comparison between lower extremities.

REVISTA
ESPAÑOLA DE PODOLOGÍA
Publicación Oficial del Consejo General de Colegios Oficiales de Podólogos

ORIGINAL
Bilingual article English/Spanish

Rev Esp Podol. 2021;xx(x):x-x
DOI: 10.20986/revesppod.2021.1594/2020

Descriptive analysis of 3D foot motion using inertial sensors: comparison between lower extremities

Análisis descriptivo del movimiento 3D del pie mediante sensores inerciales: comparación entre extremidades

Juan Luis Florenciano Restoy[1], Jordi Solé-Casals[1] y Xantal Borràs Boix[2]

[1]*Grup de Tractament de Dades i Senyals. Universitat de Vic – Universitat Central de Catalunya. Barcelona. España.* [2]*Sport Performance Analysis Research Group (SPARG). Universitat de Vic - Universitat Central de Catalunya. Barcelona. España*

Keywords:

Inertial sensor, accelerometry, 3D kinematics, foot motion.

Abstract

Objectives: The objective of this study is twofold, on the one hand, it is to make a kinematic description of the movement of the foot in the three planes of the space and, on the other hand, to determine if there are differences between the lower extremities.

Patients and methods: The study is a descriptive, observational and cross-sectional one, with a sample consisting of 40 healthy adults who are also regular runners. The assessment protocol consisted of running on a treadmill at a speed of 9 km/h. Data collection was carried out during the first 20 seconds, after which the speed of the treadmill was stabilized.

Results: In the dorsi-plantar movement, no significant differences between feet were found ($p<0.37$), whereas in the pronation-supination movement and the abduction-adduction movement significant differences were found, especially in the right foot ($p < 0.002$ and $p < 0.02$ respectively). The size of the effect in the movement in the sagittal plane was found to be very small, while in the frontal and transverse planes it increased to a medium effect.

Conclusion: During running, the foot follows a logical sequence of movements. While no significant differences exist in the dorsi-plantar movements, in the pronation-supination and abduction-adduction movements the right foot was found to have a bigger range of movement than the left foot.

Palabras clave:

Sensor inercial, acelerometría, cinemática 3D, movimiento del pie.

Resumen

Objetivos: El objetivo del presente estudio es doble: por un lado, realizar una descripción cinemática del movimiento del pie en los tres planos del espacio y, por otro, determinar si existen diferencias entre ambas extremidades.

Pacientes y métodos: Se trata de un estudio descriptivo, observacional y transversal, con una muestra de 40 corredores habituales, adultos sanos. El protocolo de valoración consistió en carrera sobre cinta rodante a una velocidad de 9 km/h. La recogida de datos se realizó durante 20 segundos, después de estabilizada la velocidad de la cinta.

Resultados: En el movimiento de flexión dorsal-flexión plantar no se observan diferencias significativas entre pies ($p < 0.37$). En el movimiento de pronación-supinación y en el de adducción-abducción sí existen diferencias significativas, siendo mayor en el pie derecho ($p < 0.002$ y $p < 0.02$ respectivamente). El tamaño del efecto es muy pequeño en el movimiento en el plano sagital, mientras que en los planos frontal y transverso es un efecto mediano.

Conclusión: Durante la carrera el pie mantiene una secuencia lógica de movimiento. Mientras no existen diferencias significativas en los movimientos de flexión dorsal-flexión plantar, en los movimientos de pronación-supinación y adducción-abducción el pie derecho tiene mayor rango de movimiento que el pie izquierdo.

Received: 09-11-2020
Accepted: 22-12-2020

Correspondence:
Xantal Borràs Boix
xantal.borras@uvic.cat

INTRODUCTION

Inertial measurement units (IMU) are portable devices which, by using a combination of accelerometers, gyroscopes and magnetometers, can be used to determine kinematic patterns carried out in any environment, including the real environment of a runner, and are an alternative to the current laboratory-based research[1-3].

Due to their small size and wireless properties, IMUs allow for the movement to be studied without restrictions. Some studies have shown that they are able to detect the changes in running biomechanics with precision and can help to determine injury factors due to mechanical overload[4,5]. A variety of researchers have used accelerometers placed on the participant's sports footwear to measure changes in running; Boutaayamou et al.[6] validated the use of two accelerometers, fixed on each shoe at the level of the heel and the proximal part of the big toe, against a conventional three-dimensional (3D) optical analysis system without finding significant differences between the two methods.

Other studies have shown evidence for asymmetry of movement during walking. For instance, Mayolas et al.[7] observed an asymmetric walking behaviour, independent of the laterality of the subjects, in a child population. The results did not reveal significant bilateral differences in the general plantar pressure, but the majority of the children were found to not only apply a higher pressure in the right hindfoot rather than in the left hindfoot, but also to do so in the midfoot and left forefoot rather than in the right forefoot.

In addition, Niu et al.[8] found that plantar pressure could be used to evaluate the foot's stability. In comparison to the non-dominant side, the dominant foot was seen to be more secure when in a single-foot stance due to the higher total contact area. This was especially true in an ankle inversion stance, due to a higher antero-posterior force ratio.

Thus, the scientific interest in the analysis of the foot movement has been on the rise with the objective of reducing the risk of injury and improving running efficiency and performance. However, there is a lack of studies which attempt to describe kinematic patterns with the use of IMUs. The objective of this study is twofold. On the one hand, it is to make a kinematic description of the movement of the foot in the three planes of movement and, on the other hand, to determine if there are differences between both extremities. The null hypothesis establishes that there are no differences between feet.

PATIENTS AND METHODS

This is a descriptive, observational and transversal study approved by the University of Vic – Central University of Catalonia's Ethical Committee, following the principles of the Declaration of Helsinki.

The sample was composed by 40 male adults (ages 43 ± 13.8 years, height 175.5 ± 7.07 cm, weight 72 ± 5.5 kg) without any alterations in their locomotor system. All of them gave their written consent prior to the evaluation.

A running treadmill (BH Fitness G6414V SPORT, Álava, Spain) was used to capture various running cycles. It has been previously observed that running on a treadmill is representative of overground running[9,10].

Two IMU units, equipped with a triaxial accelerometer, gyroscope and magnetometer (MotionPod, size 31 × 21 × 15 mm and a weight of 14g, Grenoble, France), software BioVal (RM Ingénierie. Rodez, France)[11,12] and a wireless interface (2.4 GHz, transmission range of up to 30 m, ≈ 8 h of usage, sampling rate of 30 Hz) were used to collect the data. The data from the apparatus was transferred to a PC through a USB device.

Each participant warmed up on the machine by running for three minutes at 9 km/h (2.5 m/s) in order to familiarize themselves with its speed and the environment.

Once the warming-up period was completed, the athlete rested for two-minutes and the experimental procedure was explained.

The sensor was placed in the instep of the subject's footwear using Velcro and secured with adhesive tape in order to reduce the device's vibrations (Figure 1).

Each participant wore their own footwear. It has been observed that changes in midsole hardness affect lower-extremity kinematics[13]. The same footwear had to be used in the two evaluated conditions.

According to the manufacturer's protocol, the subject had to first remain still in an upright position for three seconds whilst data from both feet was being registered.

After the preparation, each participant's running was recorded for 20 seconds at the same previous speed of 9 km/h. The data began being collected once the treadmill had stabilised at the set speed.

Angular displacement data was measured between the maximum and minimum angular points in the sagittal plane (foot dorsi-plantar flexion), the transversal plane (abduction-adduction), and frontal plane (pronation-supination). Explanatory note: The manufacturer introduces the concept of pronation-supination when, in reality, based on the podiat-

Figure 1. Left: Image of the sensor used in this study. Right: Initial position of the participant on the running treadmill, with the sensor fixed on the footwear. Author's source.

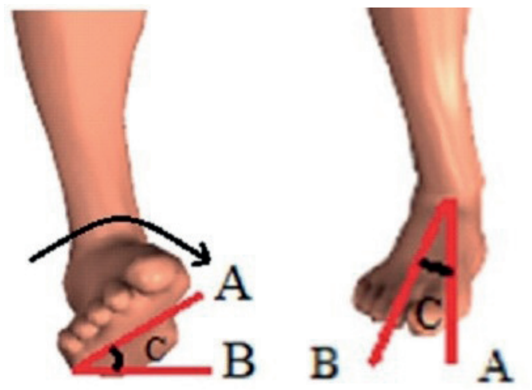


Figure 2. Left: Pronation movement of the foot. Line A indicates the supination plane, and line B indicates the plane of the ground. Right: Abduction movement of the foot. Line A indicates the sagittal plane, whilst line B indicates the foot's direction. Author's source.

ric definition, it should be inversion-eversion. This is because eversion is a movement on the frontal plane where the midfoot zone approaches the plane of the ground, and inversion is a movement on the frontal plane where the midfoot zone separates from the ground plane, whereas pronation is a triplane movement of flexion, eversion and abduction and supination is a combination of inversion, adduction and plantar flexion[14] (Figure 2).

The data was exported to an Excel spreadsheet for further analysis. Normality of the data was analysed using Shapiro-Wilk test. For the comparison of means, the Student's *t*-Test was performed with a confidence interval of 95%, recognizing those values with a p-value of less than 0.05 as being statistically significant. The effect size of the results was calculated using the Cohen test ($d \leq 0.2$ negligible, $0.2 \leq d \leq 0.5$ small, $0.5 \leq d \leq 0.8$ medium o $d \geq 0.8$ large effect)[15,16].

RESULTS

Extremities comparison

The results are displayed in Table I. The differences between the amplitude mean of the left and right feet were found to be insignificant in the flexion–extension movement ($p < 0.37$, with a d-Cohen effect being barely perceived). In contrast, the differences between the amplitude mean in the latero-medial movements (abduction–adduction) and in the pronation–supination movements were statistically significant, with p-values $p < 0.02$ and $p < 0.002$, and with medium effect sizes of 0.48 and 0.55, respectively. The higher values were observed in right foot.

Description of the curves

Figure 3 shows the angular displacement of the foot for a running cycle in the three axes of movement.

– **Dorsal flexion-Plantar flexion:** The running cycle starts on the point of maximum dorsal flexion, descending towards the X axis and drawing a slight plateau shape during the middle stance before turning into a plantar flexion. In the swing phase, the transition to a flexion is produced.
– **Abduction-adduction:** In the stance phase, at the touchdown point, the foot is parallel to the X axis, drawing a plateau shape during the middle stance and descending into abduction during the impulse phase. In the swing phase, the movement changes to adduction, and this is maintained throughout the phase until the touchdown (Figure 2).
– **Pronation-supination:** The touchdown begins at the point of minimum pronation of the curve, ascending towards the X-axis towards maximum pronation and drawing a plateau shape during mid-posture. The curve shows a discrete supination during the impulse phase. In the swing phase, the pronation curve changes to supination and the foot is positioned for the initiation of contact.

DISCUSSION

The first aim of this study was to kinematically describe the amplitude of the 3D movement of the left and right feet in a complete running cycle. The curve analysis has led to the observation that the foot movement follows a logical sequence; that is, in the stance phase the start of the extension of the foot coincides with the start of pronation, the foot's stabilization occurs parallel to the X axis, and in the elevation of the heel during the propulsive phase (leverage phase) the foot carries out an abduction. In the swing phase, the foot

Table I. Mean and standard deviations of angular displacement for each extremity. Values in degrees (°). *Statistically significant differences as p<0.05.

	Left foot	Right foot	Student's t-test	Cohen's d test
Dorsi-Plantar Flexion	94.9 ± 12.5	93.8 ± 13.5	p < 0.37	0.07
Pronation-Supination	16.4 ± 5.0	19.2 ± 4.8	p < 0.002*	0.55
Abduction-Adduction	22.4 ± 7.5	26.4 ± 9.0	p < 0.02*	0.48

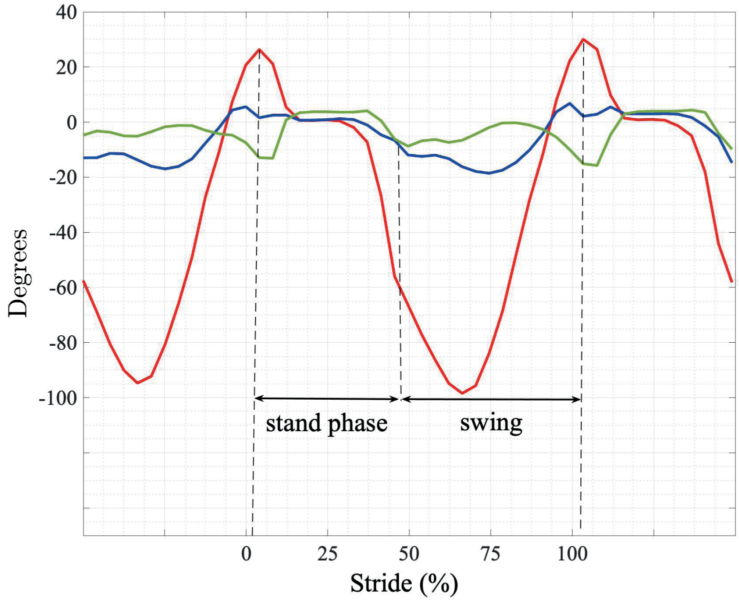


Figure 3. Graphic of the angular displacement of the foot in a running cycle. The legend on the lower left indicates the direction of the movement. Author's source.

combines flexion, inversion and adduction. Combining these 3D movements during the stance phase is crucial to establish the unipedal balance, and in the swing phase, the combination of the 3D movements place the foot correctly for the start of the new running cycle.

The second objective study was to assess whether differences between the movement of the feet were present. Although the results of the study do not reveal significant differences in the dorsi-plantar flexion, a bigger amplitude of movement has been observed in the frontal plane with an eversion movement more pronounced in the right foot. Therefore, the null hypothesis could be rejected. This could suggest that since the sagittal plane is the plane of reference, the alternation of support between the feet does not alter the mechanics, however, eversion movement is more pronounced in the right foot, meaning that it tends to have higher degrees of inversion at the start of contact than the left foot.

When it comes to the abduction movement, it is also more pronounced in the right foot, which could suggest that higher pronation implies higher abduction too, since the foot is a segment that does not execute pure movements but instead carries out combined movements.

As a matter of fact, these movements are not pure, as mentioned before, but are instead combined, since the ankle joint and the tarsal joints are mechanically associated through the subtalar axis. Their oblique projection allows for the movement of the tibia to be linked with the combination of movements from the foot. For example, in a stance phase, the internal rotation of the tibia generates a pronation movement on the foot, and similarly an external rotation of the tibia leads to a supination movement of the foot[17].

We have not found any studies that compares angular displacement mechanics between the limbs, although other studies have found differences between extremities using other types of mechanical variables. Cowley[18] analysed the change in height of the navicular bone in 30 runners (12 women and 18 men) after running 21 km, and found a significant lowering of the foot arch in both feet (4.2 mm in the left foot and 5.0 mm in the right foot). The study thus showed a change in foot posture, with a descent of the medial arch (this effect being more pronounced in the right foot) but did not explain the reasons for this change.

Stodólka et al.[19] examined the level of bilateral symmetry between the trajectory of the centre of pressure (CoP) of the left and right feet in the lateral-medial and antero-posterior directions. On the one hand, it was observed that 88% of the participants displayed symmetry in both feet for the magnitude and direction of the antero-posterior trajectory of the CoP, but on the other hand, asymmetry was observed in 67% of the participants for the latero-medial trajectory; CoP displacement was noted along the lateral limit of one foot and along the medial limit of the other. Similarly, Muntanyola[20], in a study on 663 subjects, discovered that the displacement, range and velocity of the CoP in the antero-posterior axis were bigger than in the latero-medial axis, and the majority of the subjects also showed a higher pressure on the right foot.

Rai et al.[21] registered footprints in 66 subjects, with and without a pathology, using an electronic pedobarograph. The results showed an asymmetric distribution of the plantar pressure in the right and left feet of the subjects without a pathology (17 % had the same pressure on both feet, 7 % had higher pressure on the left foot, and 76 % pressure on the right foot).

Thus, it seems that scientific evidence exists where certain values have been found to be significantly different or higher in the right foot compared to the left foot. The results presented in this study also seem to support this trend. It is of our thinking that the condition of laterality must have some influence in this. For instance, Hardyck[22] suggested that a preference on the use of the left hand, ranging from moderate to strongly left-handed, would be found on approximately only 10 % of the population. Nevertheless, a left-handed population should be studied before this link can be confirmed.

We regard that this pattern of movement should be considered to be normal, although it is true that any deviation from the averages observed in the movements in the frontal and transverse planes, would be susceptible to generate imbalances and, consequently, pathology in the locomotive system.

One limitation of our study was that we did not evaluate neither the laterality nor lateral dominance of the subjects, two different concepts following Carpes et al.[23], not being able to know if the differences observed are due to the predominance of right-handed or right-leg dominant population. Further studies are needed to assess correlation between kinematics and lateral dominance. It is also suggested to increase the number of subjects to corroborate and validate these results.

In conclusion, the results obtained in the present study did not show statistical significant differences in the range of motion between both feet in the sagittal plane, while significant differences were found in the frontal and transverse planes. Differences are more noticeably on the right right foot in a samlpe of normal healthy runners. This study show a logical kinematic pattern in the movement of the foot and, despite the asymmetry observed between limbs, the values for this running speed must be considered to be normal.

ACKNOWLEDGEMENTS

The authors world like to thank Juan Luis Florenciano Merino for this technical assistance in the preparation of this article.

CONFLICT OF INTERESTS

Authors do not have any conflict of interests.

FUNDING

None.

REFERENCES

1. Benson LC, Claremont CA, Bošnjak E, Ferber R. The use of wearable devices for walking and running gait analysis outside of the lab: a systematic review. Gait Posture. 2018;63:124-38. DOI: 10.1016/j.gaitpost.2018.04.047.
2. Reenalda J, Maartens E, Homan L, Buurke JJ. Continuous three-dimensional analysis of running mechanics during a marathon by means of inertial magnetic measurement units to objectify changes in running mechanics. Biomech J. 2016;49:3362-7. DOI: 10.1016/j.jbiomech.2016.08.032.
3. Wouda FJ, Giuberti M, Bellusci G, Maartens E, Reenalda J, Van Beijnum JF, et al. Estimation of vertical ground reaction forces and sagittal knee kinematics during running using three inertial sensors. Front Physiol. 2018;9:1-14. DOI: 10.3389/fphys.2018.00218.
4. Kiernan D, Hawkins DA, Manoukian MAC, McKallip M, Oelsner L, Caskey CF, et al. Accelerometer-based prediction of running injury in National Collegiate Athletic Association track athletes. Biomech. 2018;73:201-9. DOI: 10.1016/j.jbiomech.2018.04.001.
5. Schütte K H, Seerden S, Venter R, Vanwanseele B. Influence of outdoor running fatigue and medial tibial stress syndrome on accelerometer-based loading and stability. Gait Posture. 2018;59:222-8. DOI: 10.1016/j.gaitpost.2017.10.021.
6. Boutaayamou M, Schwartz C, Stamatakis J, Denoël V, Maquet D, Forthomme B, et al. Development and validation of an accelerometer-based method for quantifying gait events. Med Eng Phys. 2015;37:226-32. DOI: 10.1016/j.medengphy.2015.01.001.
7. Mayolas Pi C, Legaz Arrese A. Distribution of plantar pressures during gait in different zones of the foot in healthy children: the effects of laterality. Percept Mot Skills. 2015;120(1):159-76. DOI: 10.2466/26.10.PMS.120v15x8.
8. Niu WX, Chu ZW, Yao J, Zhang M, Fan YB, Zhao QP. Effects of laterality, ankle inversion and stabilizers on the plantar pressure distribution during unipedal standing. J Mech Med Biol. 2012;12(3):1250055. DOI: 10.1142/S0219519412005113.
9. Fellin RE, Manal K, Davis IS. Comparison of lower extremity kinematic curves during overground and treadmill running. J Appl Biomech. 2010;26:407-14. DOI: 10.1123/jab.26.4.407.
10. Fellin R, Rose M, William C, Royer D, Todd D. Comparison of methods for kinematic identification of footstrike and toe-off during overground and treadmill running. J Sci Med Sport. 2010;13:646-50. DOI: 10.1016/j.jsams.2010.03.006.
11. MPOD1. MotionPod Manual de usuario. Movea Motion copyright © 2008-2009.
12. Manual de Instrucciones. BioVal - RM Ingenierie Movea Motion copyright © 2008-2009.
13. Hardin EC, Van Den Bogert AJ, Hamill J. Kinematic adaptations during running: effects of footwear, surface, and duration. Med Sci Sports Exerc. 2004;36(5):838-44. DOI: 10.1249/01.MSS.0000126605.65966.40.
14. Root ML, Orien WP, Weed JH. Biomechanical Examination of the Foot. Vol. 1. Editorial Ortoce; 1991.
15. Cohen J. Statistical power analysis for the behavioral sciences. 2ª ed. Hillsdale NJ: Lawrence Erlbaum Associates, Publishers; 1988.
16. Chuan CL, Penyelidikan J. Sample size estimation using Krejcie and Morgan and Cohen statistical power analysis: A comparison. Jurnal Penyelidikan IPBL. 2006;7:78-86.
17. Kapandji IA. Fisiología articular. Miembro inferior. 6ª ed. Editorial Maloine; 2012.
18. Cowley E. The effects of prolonged running on the biomechanics and function of the foot and ankle. A thesis submitted to the University of Plymouth in partial fulfilment for the degree of School of Health Professions. September, 2019.
19. Stodółka J, Blach W, Vodicar J, Máckała K. The characteristics of feet center of pressure trajectory during quiet standing. Appl Sci. 2020;10(8):2940. DOI: 10.3390/app10082940.
20. Muntanyola-Vidal A. Medida del equilibrio estático en corredores de maratón mediante baropodometría optométrica. Universitat Ramón Llull. Tesis doctoral. Barcelona. Mayo, 2014.
21. Rai DV, Aggarwal LM, Bahadur Raj. Plantar pressure changes in normal and pathological foot during bipedal standing. Indian J Orthop. 2006;40(2):119-22. DOI: 10.4103/0019-5413.34455.
22. Hardyck C, Petrinovich F. Left-handedness. Psychol Bull. 1997;84:385-404. DOI: 10.1037/0033-2909.84.3.385.
23. Carpes FP, Mota CB, Faria IE. (2010). On the bilateral asymmetry during running and cycling–A review considering leg preference. Phys Ther Sport. 2010;11(4):136-42. DOI: 10.1016/j.ptsp.2010.06.005.

5.2. ESTUDIO 2: IMU-based effects assessment of the use of foot orthoses in the stance phase during running and asymmetry between extremities.

sensors

MDPI

Article

IMU-Based Effects Assessment of the Use of Foot Orthoses in the Stance Phase during Running and Asymmetry between Extremities

Juan Luis Florenciano Restoy [1], Jordi Solé-Casals [1,*] and Xantal Borràs-Boix [2]

[1] Data and Signal Processing Research Group, University of Vic–Central University of Catalonia, 08500 Vic, Spain; juanluis.florenciano@uvic.cat

[2] Sport Performance Analysis Research Group, University of Vic–Central University of Catalonia, 08500 Vic, Spain; xantal.borras@uvic.cat

* Correspondence: jordi.sole@uvic.cat

Abstract: The objectives of this study were to determine the amplitude of movement differences and asymmetries between feet during the stance phase and to evaluate the effects of foot orthoses (FOs) on foot kinematics in the stance phase during running. In total, 40 males were recruited (age: 43.0 ± 13.8 years, weight: 72.0 ± 5.5 kg, height: 175.5 ± 7.0 cm). Participants ran on a running treadmill at 2.5 m/s using their own footwear, with and without the FOs. Two inertial sensors fixed on the instep of each of the participant's footwear were used. Amplitude of movement along each axis, contact time and number of steps were considered in the analysis. The results indicate that the movement in the sagittal plane is symmetric, but that it is not in the frontal and transverse planes. The right foot displayed more degrees of movement amplitude than the left foot although these differences are only significant in the abduction case. When FOs are used, a decrease in amplitude of movement in the three axes is observed, except for the dorsi-plantar flexion in the left foot and both feet combined. The contact time and the total step time show a significant increase when FOs are used, but the number of steps is not altered, suggesting that FOs do not interfere in running technique. The reduction in the amplitude of movement would indicate that FOs could be used as a preventive tool. The FOs do not influence the asymmetry of the amplitude of movement observed between feet, and this risk factor is maintained. IMU devices are useful tools to detect risk factors related to running injuries. With its use, even more personalized FOs could be manufactured.

Keywords: running; kinematics; inertial measurement unit (IMU); foot orthoses; asymmetry

Citation: Florenciano Restoy, J.L.; Solé-Casals, J.; Borràs-Boix, X. IMU-Based Effects Assessment of the Use of Foot Orthoses in the Stance Phase during Running and Asymmetry between Extremities. *Sensors* **2021**, *21*, 3277. https://doi.org/10.3390/s21093277

Academic Editor: James F. Rustling

Received: 26 March 2021
Accepted: 7 May 2021
Published: 10 May 2021

1. Introduction

Interest in the analysis of the foot strike pattern (FSP) has increased due to its association with a reduced risk of injury [1–3]. Furthermore, running gait pattern (GP) analyses can be used for injury prevention and treatment, as well as in performance enhancement [4]. GP and FSP have been analyzed using a variety of methods, such as two-dimensional (2D) video analysis, three-dimensional (3D) video analysis, center of pressure and force plate [5–8]. The 3D motion analysis system with cameras is considered to be the "gold standard", but this system is expensive and complicated to use, requiring a set of reflective markers located on the appropriate reference points to be able to calculate the desired parameters [3,5,9].

Nowadays, there is a growing interest in evaluating GP and FSP in environments outside of the laboratory, and inertial measurement units (IMUs) are an interesting option to do so due to their reduced size and wireless properties [10]. An IMU sensor is the wearable device that will be used in our experiments to calculate the dorsi–plantar flexion (D–PF), the abduction–adduction (ABD–ADD) and the eversion–inversion (EV–INV) movements of the foot, using a combination of data from a triaxial accelerometer, a gyroscope and a triaxial magnetometer. The IMU sensor devices have attractive features such as low cost, low power consumption and overall simplicity in their use, and allow for the continuous monitoring of the subject's daily activities, such as walking or running, either on a treadmill, or in real environments [10,11], but they also suffer from some disadvantages such as interference or measurement errors that must be taken into account. IMU devices have been used to determine gait kinematic changes when running [3,11–19], and several research studies have evaluated the reliability of IMU devices. For example, Boutaayamou et al. [12] validated the use of accelerometers, fixed on each shoe at the level of the heel and the proximal part of the big toe, against a 3D optical analysis system. Giandolini et al. [13] compared the use of accelerometers, placed on the external faces of the shoes at the heel and metatarsal levels, with the use of 2D video analysis. They compared the time between heel and metatarsal accelerations, as well as the foot strike angle, and it was determined that the method is reliable for a wide range of velocities. Sinclair et al. [17] concluded that accelerometers placed on the distal tibia can be used to precisely detect events during walking. Finally, Shiang et al. [19] used two IMUs (one accelerometer and one gyroscope) on the upper part of the shoe to determine the foot strike angle during running and found a significant correlation between the strike angle and the sagittal plane angles, which were acquired from a 3D system.

Running is the natural evolution of gait when increasing the velocity of movement. A stride can be divided into a stance phase, which occurs between foot strike and toe off, and a swing phase, which includes the double float, and the stance phase of the contralateral limb [4]. Running is one of the most popular physical activities, but it is also an activity with relatively high injury rate, with an estimation of one injury per year in 50% of all runners [20]. It is well known that injury occurrence is multifactorial [21], and, in running, there are several predisposing risk factors including: increased vertical ground reaction force relative to walking (2.2 times body weight after heel contact in running compared to 1.1 times body weight during walking) [4], long running distances, history of previous injury, cavus feet, muscle weakness, excessive supination during stance phase and bilateral asymmetry [20–23]. Previous references have highlighted that having a limb asymmetry greater than 15% is associated with an increased incidence of injury in both athlete and non-athlete populations [24,25].

Athletic foot orthoses (FOs) are shoe inserts that replace the removable stock insole. They have proven to be effective as a treatment for sport-related injuries and health diseases [20,26–29]. For example, Mündermann et al. [27] ascertained that the use of FOs reduced the maximum foot eversion and the ankle inversion moment, modifying the vertical loading rate and the maximum knee external rotation moment, and Lack et al. [28], through the placing of anti-pronation FOs, observed a significant decrease in hip adduction and in the internal rotation of the knee after the foot strike in subjects with severe pronation. Furthermore, Brognara et al. [29] observed that foot plantar stimulation using a 3D-printing insole generated more stable walking pattern in Parkinson's disease patients.

It has also been observed that FOs could be a part of the armor in the prevention of running injuries [20,30,31]. For example, Franklyn-Miller et al. [31] evaluated foot pressures in four hundred aspiring military officers when running using pressure insoles. Out of the two groups, the subjects in the group who used personalized foot orthoses reduced overall injuries and stress fractures but not soft-tissue injuries. All of the studies referred analyzed parameters that are produced in the stance phase of the gait, so this is the phase chosen for analysis in the present study.

Healthy gait is assumed to be symmetrical, but asymmetries often exist [23,32]. Vagenas and Hoshizak [23] suggested that running shoes could significantly decrease the degree of rearfoot asymmetry, but no study has been found on the effect of FOs over the asymmetry of running.

Most articles focus on analyzing the effect of FOs from the perspective of injury [20,26–29], but this article does so from the perspective of non-injury and, therefore, prevention. The aim of this study is twofold: (1) to determine kinematic differences and asymmetries between feet during stance phase; (2) to evaluate the effects of FOs on foot kinematics. For the first objective, it is hypothesized that the kinematics of the foot will be symmetrical, and for the second objective, it is hypothesized that there will be no differences in foot motion when FOs are used.

As far as we know, this kind of study has never been conducted before with such a considerable number of subjects. Hence, the obtained results are highly useful and will allow us to have a deeper knowledge of the GP and the real effects of FOs in the stance phase of the stride.

2. Materials and Methods

This is a quantitative observational study following STROBE methodology [33].

The sample was composed by 40 male adult participants (age 43.0 ± 13.8 years; height 175.5 ± 7.0 cm; weight 72.0 ± 5.5 kg), regular amateur runners, that had been using FOs for at least one year. None of the subjects presented any alterations in their locomotor system.

All the subjects were informed of the conditions of the study and signed an informed consent prior to their participation. All the tests were non-invasive and followed the principles of the Declaration of Helsinki [34]. The study was approved by the University of Vic—Central University of Catalonia's Ethical Committee (UVic-UCC, 09/2016).

2.1. Data Acquisition Procedure

Two IMUs, equipped with a triaxial accelerometer, a gyroscope and magnetometer (MotionPod, sampling rate of 30 Hz, size 33 × 22 × 15 mm, a weight of 14 g, Grenoble, France), along with a wireless interface (2.4 GHz, transmission range of up to 30 m, ≈8 h of usage, Grenoble, France) were used to collect the data. The software used was Logiciel Medical (RM Ingenierie, Rodez, France) [35,36]. This is a commonly used IMU in podiatry and physical therapy, with factory default settings. It has been shown to have the potential to assess movement and coordination variability between and within individuals from joint angle measures in swimming and limb orientation time-series data in climbing [37].

The analysis procedure started with the placing of the sensor on the instep of the subject's sports footwear, using Velcro, which was then secured with adhesive tape to reduce vibrations in the device. Each participant brought their own footwear that they used regularly for running. It has been observed that changes in the hardness of the sole of the shoe can affect the kinematics of the inferior extremities [38,39]. The same footwear had to be used in the two evaluated conditions.

Each participant carried out a warm-up run on the running treadmill (BH Fitness G6414V SPORT, Álava, Spain) by running for three minutes at 2.5 m/s (9 km/h) without any inclination (measured with a bubble level) to familiarize themselves with the machine speed and the environment. It has been previously observed that running on a treadmill is representative of overground running [40,41]. Once the warm-up period had been completed, the participant rested for two minutes, and the rest of the experimental procedure was then explained.

According to the manufacturer's protocol and instructions, the participant first remained still and upright for 3 s while the sensor was being calibrated. Thus, in a triple orthogonal relative reference system, the vertical axis collects the ABD–ADD movements of the foot, the longitudinal axis collects the EV–INV movements, and the D–PF movements are collected in the medio-lateral axis (Figure 1).

Considering that the transition between walking and running occurs at approximately 2.2 m/s [4], a low running speed of 2.5 m/s was chosen to meet the IMU sampling rate requirements. Consequently, two runs at 2.5 m/s were carried out, the first one with the subjects using their regular sports footwear and the second one wearing their regular sports footwear and FOs. As the effects of the FOs are mechanical and the acquisition time of the IMUs was 20 s, it was decided not to randomize the tests as it is considered that there is no familiarization effect in the run. Furthermore, we must take into account that the runners included, although recreational, have experience and accumulated mileage, and are already familiar with the FOs.

Figure 1. (**Left**): position of the sensor attached to the instep of the running shoe. The triple orthogonal system represented by the arrows indicate the dorsi–planar flexion (red), abduction–adduction (blue) and eversion–inversion (green) movements. (**Right**): type of FOs used with the polypro-pylene and the EVA layers.

Following the IMU manufacturer's recommendations, data acquisition was performed for 20 s after the treadmill speed stabilized. This time is guaranteed by MotionPod to ensure data reliability and avoid measurement errors caused by data integration over time [35]. The initial steps when transitioning between stationary and running phases, as well as the deceleration steps at the end, were discarded.

The manufacturing system of the FOs was that of thermoforming, with adaptation through a vacuum chamber. In the structure of the foot support, the thermoplastic material Polypropylene (3 mm) was used, and the cushioning material placed on the Polypropylene was Ethylene-Vinyl Acetate (EVA) of Shore 30 hardness (Figure 1).

2.2. Data Processing

The data was processed using Matlab (Mathworks, Natick, MA, USA). Taking the movement in the sagittal plane (D–PF) as a reference, the stance phase was determined for each step, which lies between the points of maximum and minimum plantar flexion. Thus, the starting and ending points of the stance phase of each step in the frontal and transverse planes are also obtained.

Subsequently, a mathematical adjustment was carried out to ensure that in the D–PF, the plateau zone (mid-stance) corresponded with the 0° of the movement, and the possible deviations of the sensor in the movements of EV–INV and ABD–ADD were also corrected, adjusting the start at 0°. The swing phase was discarded in this study (Figure 2).

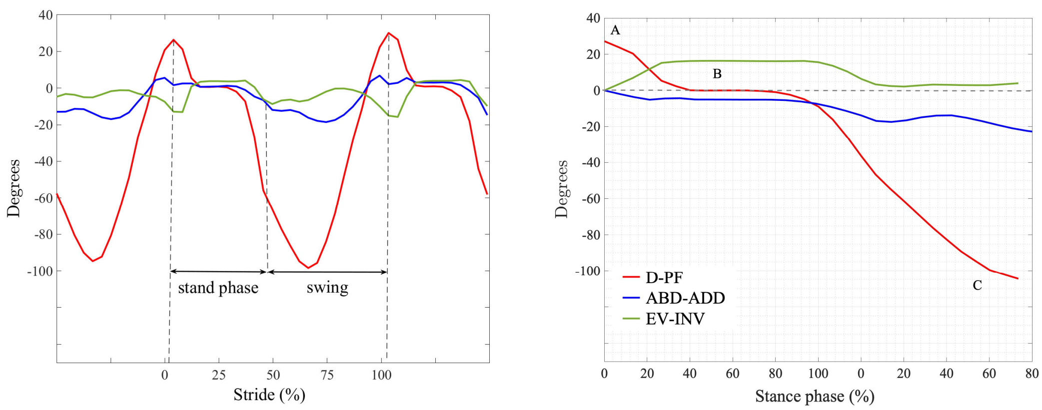


Figure 2. (**Left**): angular displacement of the foot as a function of the percentage of the running cycle. (**Right**): stance phase segmentation graph where A indicates the start of contact, B corresponds to the mid-stance (stabilization), and C indicates the end of the stance. Legend: D–PF indicates dorsi–plantar flexion, ABD–ADD indicates abduction–abduction, EV–INV indicates eversion–inversion.

2.3. Data Analysis

The mean amplitude of maximum movement for each one of the axes and conditions was determined for the data analysis through the following functions:

$$\overline{f}(t) = \frac{\sum_{i=1}^{N} f_i(t)}{N},\ \overline{a}(t) = \frac{\sum_{i=1}^{N} a_i(t)}{N},\ \overline{p}(t) = \frac{\sum_{i=1}^{N} p_i(t)}{N} \quad (1)$$

where $f_i(t)$, $a_i(t)$ and $p_i(t)$ are the functions representing the time evolution of the movement for D–PF, ABD–ADD and EV–INV movements, respectively; N refers to the number of steps taken by each subject during the 20 s of data collection.

Moreover, in the analysis between extremities, the asymmetry percentage was analyzed using Equation (2) [42]. This equation is considered to be accurate for the calculations of asymmetries in unilateral tests [43].

$$Asymmetry = -100\,\frac{minimum\ value}{maximum\ value} + 100 \quad (2)$$

Matlab was used for the statistical analysis. Data normality was analyzed through the Kolmogorof–Smirnov test was assessed before doing the main test. The differences between the conditions were evaluated using a paired sample mean test (T-Student), confidence interval and the level of significant alpha were set at 95% and 0.05, respectively. To allow for a better interpretation of the data, the effect size (d-Cohen) was carried out with the following criteria: $d \leq 0.2$ negligible, $0.2 \leq d \leq 0.5$ small effect, $0.5 \leq d \leq 0.8$ medium effect, $0.8 \leq d$ large effect [44–46].

3. Results

3.1. Comparison between Feet

When comparing the kinematics between the feet (Table 1), it can be observed that the left and right feet present, on average, the same degrees of amplitude of movement in the sagittal plane (D–PF); in tenths of degrees, the right foot displays a smaller amplitude, although this difference is not statistically significant (p = 0.76), and the size of the effect is practically negligible (d = 0.05). In the frontal plane (EV–INV) and the transverse plane (ABD–ADD), the right foot exhibits slightly larger amplitude of movement values. These differences are not statistically significant for EV–INV (p = 0.09), but they are significant for ABD–ADD (p = 0.03). The size of the effect is small for both movements (d-EV–INV = 0.28; d-ABD–ADD = 0.37).

Table 1. Means and standard deviations of the angular displacement of each one of the extremities for each axis of movement.

	Left	Right	Asymmetry	p-Value	d-Cohen
Dorsal–Plantar Flexion	96.6 ± 13.5	96.2 ± 14.7	6.2 ± 5.2	0.76	0.05
Eversion–Inversion	12.2 ± 3.8	13.36 ± 3.9	31.9 ± 18.2	0.09	0.28
Abduction–Adduction	21.3 ± 7.4	25.4 ± 9.8	21.7 ± 14.5	0.03 *	0.37

Values for the angular displacement for each left and right leg in degrees. Values of asymmetry as a percentage. * Indicates statistically significant differences between angular values in the right and left legs (p < 0.05).

3.2. Comparison between Footwear with and without FOs

The data relating to the amplitude of movement, the time and the number of steps are presented in Table 2. The values are separated between the right foot, left foot and both feet.

Table 2. Angular displacement, time variables and number of steps in each of the extremities (individually and for both feet) for each axis of movement.

		Footwear	Orthoses (FOs)	Difference	*p*-Value	d-Cohen
Left	Dorsi–plantar flexion (°)	96.6 ± 13.5	97.2 ± 12.1	0.62 ↑	0.42	−0.13
	Eversion–Inversion (°)	12.2 ± 3.8	11.5 ± 3.8	−0.77 ↓	0.02 *	0.37
	Abduction–Adduction (°)	21.3 ± 7.4	19.5 ± 6.9	−1.76 ↓	0.06	0.30
	Contact time (ms)	454 ± 41	459 ± 39	5 ↑	0.01 *	−0.42
	Total time (ms)	730 ± 54	736 ± 53	6 ↑	0.01 *	−0.40
	Number of steps	14.9 ± 1.5	14.8 ± 1.4	0.1 =	0.44	0.12
Right	Dorsi–plantar flexion (°)	96.2 ± 14.7	96.1 ± 13.5	−0.06 ↓	0.94	0.01
	Eversion–Inversion (°)	13.4 ± 3.9	13.2 ± 3.9	−0.17 ↓	0.75	0.05
	Abduction–Adduction (°)	25.4 ± 9.8	23.4 ± 8.9	−2.01 ↓	0.05	0.32
	Contact time (ms)	455 ± 42	461 ± 40	6 ↑	0.01 *	−0.47
	Total time (ms)	730 ± 55	736 ± 53	6 ↑	0.01 *	−0.39
	Number of steps	15.2 ± 1,3	15.0 ± 1.3	0.2 =	0.04 *	0,34
Both	Dorsi–plantar flexion (°)	96.4 ± 14.2	96. 7 ± 12.9	0.28 ↑	0.60	−0.06
	Eversion–Inversion (°)	12.8 ± 3.9	12.3 ± 4.0	0.47 ↓	0.13	0.17
	Abduction–Adduction (°)	23.3 ± 9.0	21.5 ± 8.2	1.89 ↓	0.01 *	0.31
	Contact time (ms)	454 ± 42	460 ± 40	6 ↑	<0.01 *	−0.45
	Total time (ms)	730 ± 55	736 ± 53	6 ↑	<0.01 *	−0.40
	Number of steps	15.0 ± 1.4	14.9 + 1.3	0.1 =	0.05	0.22

Mean and standard deviation for the variables. * Indicates statistically significant differences ($p < 0.05$).

It can be observed that the FOs decrease the values for the movement amplitudes, with the exception of the D–PF in the left foot and in both feet, which increase by a few tenths of a degree. This reduction in the amplitude of movement is statistically significant for the EV–INV in the left foot ($p = 0.02$) and for the ABD–ADD in both feet ($p = 0.01$). The size of the effect of the FOs is either negligible or small in both cases.

The differences in the time variables lie between 5 and 6 ms, significantly incrementing when the FOs are used. The number of steps is kept at a stable value for footwear with or without FOs.

The asymmetry between the extremities when FOs are used is of 5.3 ± 0.8% for D–PF, 24.8 ± 16.1% for EV–INV and 34.8 ± 18.2% for ABD–ADD. Here, no statistically significant differences are present in any case with the condition of the footwear without FOs ($p = 0.07$ for D–PF, $p = 0.21$ for EV–INV, $p = 0.34$ for ABD–ADD).

4. Discussion

4.1. Comparison between Feet

The null hypothesis established that there would be no differences between the feet in terms of the mean amplitude of movement.

The results (Table 1) indicate that the movement in the sagittal plane is symmetric, but not in the frontal and transverse planes. The right foot presents more degrees of amplitude of movement than the left foot in both IN–EVE and ABD–ADD, although these differences are statistically significant in the ABD–ADD case. These differences in foot kinematics are evidenced by the asymmetry percentages of IN–EVE and ABD–ADD of 21.7% and 31.9%, respectively.

Previous references have highlighted that presenting asymmetry between extremities that are bigger than 15% is associated with a higher incidence of injury, both in the athlete and non-athlete populations [24,25]. Then, the D–FP movement shows an asymmetry between extremities (6.2%) that is within the normal values, but the ABD–ADD and EV–INV show asymmetries that would represent a possible injury risk factor.

It is necessary to understand that the movements of the foot are not pure but combined. Thus, the pronation movement is carried out on three planes simultaneously: dorsal flexion, eversion and abduction. Similarly, the supination movement is carried out on the plantar flexion, inversion and adduction planes [47]. It has been observed that the eversion movement of the foot is present from the start of the stance phase until the beginning of the mid-stance, whereas the abduction movement is present from the mid-stance phase until the beginning of the swing phase. The start of the plantar flexion curve coincides with the start of the eversion and abduction curves. However, at the start of the contact, the plantar flexion and eversion curves present a larger slope than the abduction curve does, indicating that the movement is executed more quickly and that, therefore, the degree of eversion is directly proportional to the degrees of inversion before the start of the contact. Consequently, a bigger eversion is equivalent to a bigger abduction.

We have not found a study that compares the mechanics between extremities through the evaluation of the angular displacement. Nonetheless, other studies have also found differences between extremities using other types of mechanical variables. For example, Polk et al. [32] obtained gait asymmetries through the analysis of ground reaction force (GRF). They observed that vertical GRFs were very symmetrical, whereas there were significant asymmetries in the maximum mediolateral forces and impulses towards the dominant right limbs. In our study, the differences in lateromedial movements are also greater in the right foot, but we did not evaluate the laterality nor the lateral dominance of the subjects, which are two different concepts Carpes et al. [48] As a result, it cannot be known whether the observed differences are due to the predominance of the hand or to the dominant leg.

Cowley [49] analyzed the change in height of the medial foot arch after a 21 km run in 30 runners (12 women and 18 men), taking the navicular bone as a reference, and found a significant decrease in the foot arch in both feet (4.2 mm in the left foot, 5.0 mm in the right foot). Therefore, the study showed a change in posture of the foot, with a decrease in the medial arch, which was, again, more pronounced in the right foot, but did not give reasons for these changes. Stodółka et al. [50] examined the level of bilateral symmetry between the trajectory of the center of pressure (CoP) of the right and left feet in the latero-medial and antero-posterior directions. On the one hand, it was observed that 88% of the participants displayed symmetry of the left and right foot for the magnitude and direction of the antero-posterior trajectory of the CoP, but on the other hand, asymmetry was observed in 67% of the participants for the latero-medial trajectory; CoP displacement was noted along the lateral limit of one foot and along the medial limit of the other. Similarly, Montañola [51] discovered, in a study on 663 subjects, that the displacement, range and velocity of the CoP in the antero-posterior axis were bigger than in the latero-medial axis, and most of the subjects also showed a higher pressure on the right foot. On the other hand, De Carvalho et al. [52] obtained larger pronation values in the left foot than in the right foot in a male population using the foot posture index (FPI-6).

Rai et al. [53] registered footprints in 66 subjects, with and without a pathology, using an electronic pedobarograph. The results showed an asymmetric distribution of the plantar pressure in the right and left feet of the subjects without a pathology (17% had the same pressure on both feet, 7% had higher pressure on the left foot and 76% had higher pressure on the right foot).

In the rehabilitation of injuries, it is common to use the values of the contralateral leg as reference values. Nevertheless, data shown in the present study indicate that asymmetries in the kinematics of movement in non-injured people exist, supporting the results presented by Vagenas and Hoshiza [23] and Polk et al. [32]. Radzak et al. [54] also observed asymmetry in the angular values of the ankle, knee and hip in healthy subjects, as well as Gao et al. [55], who found asymmetry in the plantar pressure of the dominant extremity (the right extremity in all the subjects).

In summary, the asymmetry in the kinematic variables between the extremities during running can be seen in healthy subjects, which means that quantifying the possible differences and asymmetries could have implications in preventing injuries during walking, running and in the design choice of footwear.

It has been suggested that sidewalk running has greater kinematic variability than treadmill running [10], so it would be interesting to see, in further studies, whether these results hold outdoors in less cyclical and more variable conditions or whether they are even amplified. The technology used in this study could also be used to assess this, although a higher frequency of capture would be required.

4.2. Comparison between Footwear with and without FOs

The null hypothesis established that there would not be differences in the movement of the foot in favor of the FOs.

The results indicate (Table 2) that FOs decrease the amplitude of movement in the three axes, except for the D–PF in the left foot and both feet combined. For the D–PF movement, the differences are neither statistically significant in any of the two feet nor are they statistically significant when analyzing both feet combined. These results should be considered when talking about runners with functional hallux limitus as the FOs would further reduce the dorsiflexion of the first metatarsal. The less this movement, the more the foot is forced to take off in adduction [56].

The data obtained suggest that the FOs reduce the amplitude of movement of the EV–INV and ABD–ADD movements in both feet, although there is no significant difference agreement. If the number of subjects were to be increased, higher differences in these amplitudes of movement could probably be found. It is suggested that tendency is talked about instead of definite trends. Excessive supination during the stance phase has been observed as a risk factor for running injuries [20]. Therefore, a reduction in the amplitude of movement would reduce supination indicating that FOs could be used as a preventive tool. This is in accordance with Franklyn-Miller et al. [31] who found that the use of the personalized FOs reduced the frequency of injuries in a group of military officials undergoing training.

These results are in line with those observed in other studies [27–29,57,58]. For instance, Mündermann et al. [27] observed that the use of the FOs reduced the maximum foot eversion, while Lack et al. [28] observed a reduction in the hip adduction and in the internal rotation of the knee after foot strike in patients with severe pronation. Nawoczenski et al. [57] found that the use of FOs reduced the internal rotation of the tibia in the transverse plane.

While the present study did not evaluate the internal rotation of the tibia, we know that the tarsal joints (subtalar and transverse tarsal) connect with the tibia through the subtalar axis [58], which implies that the eversion of the foot (mid-stance phase) leads to internal rotation of the tibia. If we consider that in the mid-stance phase the action of the internal retro-malleolar muscles (especially the tibialis posterior) and the external retro-malleolar muscles (especially the peroneus longus) is eccentric, and that these muscles are responsible for the stabilization of the ankle in eversion and abduction, then the observed reduction could suggest a decrease in the activity and an increase in the efficiency of these muscles. Consequently, we suggest that by controlling this eversion mechanism, the internal rotation of the tibia could also be reduced. Moreover, the abduction movement is mainly produced in the impulse phase; therefore, reducing abduction would facilitate the impulse phase and with it, concentric muscle contraction of the aforementioned retro-malleolar muscles.

It has also been found in the literature that the use of FOs modifies other variables which have not been analyzed in this study. For example, Açak [59] compared the efficiency of personalized vs. prefabricated FOs and concluded that the individually designed FOs had a beneficial role in the normalization of the forces acting on the foot and improved the physical performance parameters of people with flat feet.

In the present study, the type of footprint (neutral, pronation and supination) of the subjects was not addressed, potentially masking, for example, a larger difference for the subjects with flat feet as Açak suggests [59]. Future studies could explore this issue.

The contact time and the total stride time present a significant increase when FOs are used. The differences, however, are, on average, 6 ms and do not affect the step frequency since the number of steps of the subjects analyzed did not change significantly with or without the use of the orthoses. These data suggest that the step length, which is directly proportional to the running velocity and inversely proportional to the step frequency, was also not modified. This is interesting, as even though FOs can modify the intrinsic kinematics of the foot, they do not interfere with the running technique.

Regarding asymmetry, it has been found that the kinematic changes occurred in both feet and that FOs had no effect on minimizing this risk factor. This is something novel which has not been described so far in the scientific literature. It is true that by varying the mechanics of the foot, the mechanics of the ankle, knee and hip are also affected [27–30,57,59], but this opens a new line of research in the fields of podiatry and sport medicine, which could lead to reducing the asymmetries found by personalizing the FOs even more.

5. Conclusions and Future Work

Based on the results of this study, we can conclude that (1) asymmetry is observed between the feet in the frontal and transverse planes, showing that the right foot has more degrees of amplitude of movement than the left foot. (2) There is a tendency for the FOs to reduce the amplitude of movement in the frontal and transverse planes, but not in the sagittal plane. (3) The kinematic changes observed using the FOs did not interfere with the technique of the run. (4) FOs do not influence the asymmetry of the amplitude of movement observed between the extremities, as the kinematic changes are produced in both legs.

IMU sensors are a good alternative for the study of locomotor system movement and can provide valuable data in a simple way. Moreover, IMU devices are useful tools to detect risk factors related to running injuries. Their generalized use would allow the manufacture of even more personalized and individualized FOs.

We are already now working on the collection of new data to expand our current sample of data and to determine if the preliminary results found can be confirmed. These new data are being collected for all the subjects and are related to their laterality and lateral dominance to be able to go more in depth in the subsequent analysis of the data.

Author Contributions: Conceptualization, X.B.-B., and J.S.-C.; methodology, X.B.-B., and J.S.-C.; software, J.L.F.R.; formal analysis, J.L.F.R.; investigation, J.L.F.R., and X.B.-B.; data curation, J.L.F.R., and X.B.-B.; writing—original draft preparation, J.L.F.R.; writing—review and editing, X.B.-B., and J.S.-C.; supervision, X.B.-B., and J.S.-C.; project administration, X.B.-B., and J.S.-C. All authors have read and agreed to the published version of the manuscript.

Funding: This research received no external funding.

Institutional Review Board Statement: The study was conducted according to the guidelines of the Declaration of Helsinki and approved by the Institutional Ethics Committee of the University of Vic—Central University of Catalonia (UVic-UCC, 09/2016).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The datasets generated and analyzed during the current study are not publicly available due to ethics and privacy requirements but are available from the corresponding author on reasonable request.

Acknowledgments: The authors would like to acknowledge Juan Luis Florenciano Merino for his technical collaboration and Pau Solé-Vilaró for the English review.

Conflicts of Interest: The authors declare no conflict of interest.

References

1. Ardigò, L.P.; Lafortuna, C.; Minetti, A.E.; Mognoni, P.; Saibene, F. Metabolic and mechanical aspects of foot landing type, forefoot and rearfoot strike, in human running. *Acta Physiol. Scand.* **1995**, *155*, 17–22. [CrossRef] [PubMed]
2. Ogueta-Alday, A.; Rodríguez-Marroyo, J.A.; García-López, J. Rearfoot striking runners are more economical than midfoot strikers. *Med. Sci. Sports Exerc.* **2014**, *46*, 580–585. [CrossRef] [PubMed]
3. Williams, D.S.; McClay, I.S.; Manal, K.T. Lower extremity mechanics in runners with a converted forefoot strike pattern. *J. Appl. Biomech.* **2000**, *16*, 210–218. [CrossRef]

4. Dugan, S.A.; Bhat, K.P. Biomechanics and analysis of running gait. *Phys. Med. Rehabil. Clin.* **2005**, *16*, 603–621. [CrossRef] [PubMed]
5. Altman, A.R.; Davis, I.S. A kinematic method for footstrike pattern detection in barefoot and shod runners. *Gait Posture* **2012**, *35*, 298–300. [CrossRef] [PubMed]
6. Cavanagh, P.R.; Lafortune, M.A. Ground reaction forces in distance running. *J. Biomech.* **1980**, *13*, 397–406. [CrossRef]
7. Hasegawa, H.; Yamauchi, T.; Kraemer, W.J. Foot strike patterns of runners at the 15-km point during an elite-level half marathon. *J. Strength Cond. Res.* **2007**, *21*, 888.
8. Williams, K.R.; Cavanagh, P.R. Relationship between distance running mechanics, running economy, and performance. *J. Appl. Physiol.* **1987**, *63*, 1236–1245. [CrossRef] [PubMed]
9. Nigg, B.M.; De Boer, R.W.; Fisher, V. A kinematic comparison of overground and treadmill running. *Med. Sci. Sports Exerc.* **1995**, *27*, 98–105. [CrossRef]
10. Benson, L.C.; Clermont, C.A.; Ferber, R. New Considerations for Collecting Biomechanical Data Using Wearable Sensors: The Effect of Different Running Environments. *Front. Bioeng. Biotechnol.* **2020**, *8*, 86. [CrossRef]
11. Bötzel, K.; Marti, F.M.; Rodríguez, M.Á.C.; Plate, A.; Vicente, A.O. Gait recording with inertial sensors–How to determine initial and terminal contact. *J. Biomech.* **2016**, *49*, 332–337. [CrossRef] [PubMed]
12. Boutaayamou, M.; Schwartz, C.; Stamatakis, J.; Denoël, V.; Maquet, D.; Forthomme, B.; Croisiera, J.-L.; Macqc, B.; Verlyb, J.G.; Garrauxd, G.; et al. Development and validation of an accelerometer-based method for quantifying gait events. *Med. Eng. Phys.* **2015**, *37*, 226–232. [CrossRef]
13. Giandolini, M.; Poupard, T.; Gimenez, P.; Horvais, N.; Millet, G.Y.; Morin, J.B.; Samozino, P. A simple field method to identify foot strike pattern during running. *J. Biomech.* **2014**, *47*, 1588–1593. [CrossRef] [PubMed]
14. Laughton, C.A.; Davis, I.M.; Hamill, J. Effect of strike pattern and orthotic intervention on tibial shock during running. *J. Appl. Biomech.* **2003**, *19*, 153–168. [CrossRef]
15. Moon, Y.; McGinnis, R.S.; Seagers, K.; Motl, R.W.; Sheth, N.; Wright, J.A., Jr.; Ghaffari, R.; Sosnoff, J.J. Monitoring gait in multiple sclerosis with novel wearable motion sensors. *PLoS ONE* **2017**, *12*, e0171346. [CrossRef] [PubMed]
16. Purcell, B.; Channells, J.; James, D.; Barrett, R. Use of accelerometers for detecting foot-ground contact time during running. In *BioMEMS and Nanotechnology II*; International Society for Optics and Photonics: Bellingham, WA, USA, 2006; Volume 6036, p. 603615.
17. Sinclair, J.; Hobbs, S.J.; Protheroe, L.; Edmundson, C.J.; Greenhalgh, A. Determination of gait events using an externally mounted shank accelerometer. *J. Appl. Biomech.* **2013**, *29*, 118–122. [CrossRef]
18. Phinyomark, A.; Osis, S.; Hettinga, B.A.; Ferber, R. Kinematic gait patterns in healthy runners: A hierarchical cluster analysis. *J. Biomech.* **2015**, *48*, 3897–3904. [CrossRef]
19. Shiang, T.Y.; Hsieh, T.Y.; Lee, Y.S.; Wu, C.C.; Yu, M.C.; Mei, C.H.; Tai, I.H. Determine the foot strike pattern using inertial sensors. *J. Sens.* **2016**, *2016*, 4759626. [CrossRef]
20. Fields, K.B.; Sykes, J.C.; Walker, K.M.; Jackson, J.C. Prevention of running injuries. *Curr. Sports Med. Rep.* **2010**, *9*, 176–182. [CrossRef]
21. Meeuwisse, W.H. Assessing causation in sport injury: A multifactorial model. *Clin. J. Sport Med.* **1994**, *4*, 166–170. [CrossRef]
22. Van Gent, R.N.; Siem, D.; van Middelkoop, M.; Van Os, A.G.; Bierma-Zeinstra, S.M.A.; Koes, B.W. Incidence and determinants of lower extremity running injuries in long distance runners: A systematic review. *Br. J. Sports Med.* **2007**, *41*, 469–480. [CrossRef]
23. Vagenas, G.; Hoshizaki, B. A multivariable analysis of lower extremity kinematic asymmetry in running. *J. Appl. Biomech.* **1992**, *8*, 11–29. [CrossRef]
24. Barber, S.D.; Noyes, F.R.; Mangine, R.E.; McColskey, J.W.; Hartman, W. Quantitative assessment of functional limitations in normal and anterior cruciate ligament-deficient knees. *Clin. Orthop. Related Res.* **1990**, *255*, 204–214. [CrossRef]
25. Impellizzeri, F.M.; Rampinini, E.; Maffiuletti, N.; Marcora, S.M. A vertical jump force test for assessing bilateral strength asymmetry in athletes. *Med. Sci. Sports Exerc.* **2007**, *39*, 2044–2050. [CrossRef]
26. Kirby, K.A. Evolution of foot orthoses in sports. In *Athletic Footwear and Orthoses in Sports Medicine*; Springer: Cham, Switzerland, 2017; pp. 19–40.
27. Mündermann, A.; Nigg, B.M.; Humble, R.N.; Stefanyshyn, D.J. Foot orthotics affect lower extremity kinematics and kinetics during running. *Clin. Biomech.* **2003**, *18*, 254–262. [CrossRef]
28. Lack, S.; Barton, C.; Woledge, R.; Laupheimer, M.; Morrissey, D. The immediate effects of foot orthoses on hip and knee kinematics and muscle activity during a functional step-up task in individuals with patellofemoral pain. *Clin. Biomech.* **2014**, *29*, 1056–1062.
29. Brognara, L.; Navarro-Flores, E.; Iachemet, L.; Serra-Catalá, N.; Cauli, O. Beneficial effect of foot plantar stimulation in gait parameters in individuals with Parkinson's disease. *Brain Sci.* **2020**, *10*, 69.
30. Bonanno, D.R.; Landorf, K.B.; Munteanu, S.E.; Murley, G.S.; Menz, H.B. Effectiveness of foot orthoses and shock-absorbing insoles for the prevention of injury: A systematic review and meta-analysis. *Br. J. Sports Med.* **2017**, *51*, 86–96. [CrossRef]
31. Franklyn-Miller, A.; Wilson, C.; Bilzon, J.; McCrory, P. Foot orthoses in the prevention of injury in initial military training: A randomized controlled trial. *Am. J. Sports Med.* **2011**, *39*, 30–37. [CrossRef]
32. Polk, J.D.; Stumpf, R.M.; Rosengren, K.S. Limb dominance, foot orientation and functional asymmetry during walking gait. *Gait Posture* **2017**, *52*, 140–146. [CrossRef]

33. Vandenbroucke, J.P.; von Elm, E.; Altman, D.G.; Gøtzsche, P.C.; Mulrow, C.D.; Pocock, S.J.; Poole, C.; Schlesselman, J.J.; Egger, M.; STROBE Initiative. Strengthening the Reporting of Observational Studies in Epidemiology (STROBE): Explanation and elaboration. *Int. J. Surg.* **2014**, *12*, 1500–1524. [CrossRef] [PubMed]
34. Holt, G.R. Declaration of Helsinki-the world's document of conscience and responsibility. *South. Med. J.* **2014**, *107*, 407. [CrossRef] [PubMed]
35. MPOD1. MotionPod Manual de Usuario. Movea Motion Copyright © 2008–2009. Available online: https://fccid.io/JJ4-MPOD1/Users-Manual/Users-Manual-1206708 (accessed on 10 May 2021).
36. Manual de Instrucciones. BioVal—RM Ingenierie Movea Motion Copyright © 2008–2009. Available online: https://www.rmingenierie.net/bioval/ (accessed on 10 May 2021).
37. Seifert, L.; L'hermette, M.; Komar, J.; Orth, D.; Mell, F.; Merriaux, P.; Grenetd, P.; Caritud, Y.; Hérault, R.; Dovgalecsa, V.; et al. Pattern recognition in cyclic and discrete skills performance from inertial measurement units. *Procedia Eng.* **2014**, *72*, 196–201. [CrossRef]
38. Lussiana, T.; Fabre, N.; Hébert-Losier, K.; Mourot, L. Effect of slope and footwear on running economy and kinematics. *Scand. J. Med. Sci. Sports* **2013**, *23*, e246–e253. [CrossRef]
39. Fischer, K.M.; Willwacher, S.; Hamill, J.; Brüggemann, G.P. Tibial rotation in running: Does rearfoot adduction matter? *Gait Posture* **2017**, *51*, 188–193. [CrossRef]
40. Firminger, C.R.; Vernillo, G.; Savoldelli, A.; Stefanyshyn, D.J.; Millet, G.Y.; Edwards, W.B. Joint kinematics and ground reaction forces in overground versus treadmill graded running. *Gait Posture* **2018**, *63*, 109–113. [CrossRef]
41. Fellin, R.E.; Manal, K.; Davis, I.S. Comparison of lower extremity kinematic curves during overground and treadmill running. *J. Appl. Biomech.* **2010**, *26*, 407–414. [CrossRef]
42. Bishop, C.; Turner, A.; Maloney, S.; Lake, J.; Loturco, I.; Bromley, T.; Read, P. Drop jump asymmetry is associated with reduced sprint and change-of-direction speed performance in adult female soccer players. *Sports* **2019**, *7*, 29. [CrossRef]
43. Bishop, C.; Read, P.; Lake, J.; Chavda, S.; Turner, A. Interlimb asymmetries: Understanding how to calculate differences from bilateral and unilateral tests. *Strength Cond. J.* **2018**, *40*, 1–6. [CrossRef]
44. Cohen, J. *Statistical Power Analysis for the Behavioral Sciences*; Academic Press: Hillsdle, MN, USA, 1988.
45. Chuan, C.L.; Penyelidikan, J. Sample size estimation using Krejcie and Morgan and Cohen statistical power analysis: A comparison. *J. Penyelid. IPBL* **2006**, *7*, 78–86.
46. Razali, N.M.; Wah, Y.B. Power comparisons of shapiro-wilk, kolmogorov-smirnov, lilliefors and anderson-darling tests. *J. Stat. Model. Anal.* **2011**, *2*, 21–33.
47. Root, M.L.; Orien, W.P.; Weed, J.H. Biomechanical Examination of the Foot. *J. Am. Podiatry Assoc.* **1973**, *63*, 28–29. [CrossRef] [PubMed]
48. Carpes, F.P.; Mota, C.B.; Faria, I.E. On the bilateral asymmetry during running and cycling—A review considering leg preference. *Phys. Ther. Sport* **2010**, *11*, 136–142. [CrossRef]
49. Cowley, E. The Effects of Prolonged Running on the Biomechanics and Function of the Foot and Ankle. Ph.D. Thesis, University of Plymouth, Plymouth, UK, 2019.
50. Stodółka, J.; Blach, W.; Vodicar, J.; Maćkała, K. The Characteristics of Feet Center of Pressure Trajectory during Quiet Standing. *Appl. Sci.* **2020**, *10*, 2940. [CrossRef]
51. Montañola Vidal, A. Medida del Equilibrio Estático en Corredores de Maratón Mediante Baropodometría Optométrica. Ph.D. Thesis, Universitat Ramon Llull, Barcelona, Spain, 2014.
52. de Carvalho, B.K.G.; Penha, P.J.; Ramos, N.L.J.P.; Andrade, R.M.; Ribeiro, A.P.; João, S.M.A. Age, Sex, Body Mass Index, and Laterality in the Foot Posture of Adolescents: A Cross Sectional Study. *J. Manip. Physiol. Ther.* **2020**, *43*, 744–752. [CrossRef]
53. Rai, D.V.; Aggarwal, L.M.; Bahadur, R. Plantar pressure changes in normal and pathological foot during bipedal standing. *Indian J. Orthop.* **2006**, *40*, 119. [CrossRef]
54. Radzak, K.N.; Putnam, A.M.; Tamura, K.; Hetzler, R.K.; Stickley, C.D. Asymmetry between lower limbs during rested and fatigued state running gait in healthy individuals. *Gait Posture* **2017**, *51*, 268–274. [CrossRef]
55. Gao, Z.; Mei, Q.; Xiang, L.; Baker, J.S.; Fernandez, J.; Gu, Y. Effects of limb dominance on the symmetrical distribution of plantar loading during walking and running. *Proc. Inst. Mech. Eng. Part P J. Sports Eng. Technol.* **2020**, 1754337120960962. [CrossRef]
56. Sánchez-Gómez, R.; Becerro-de-Bengoa-Vallejo, R.; Losa-Iglesias, M.E.; Calvo-Lobo, C.; Navarro-Flores, E.; Palomo-López, P.; Romero-Morales, C.; López-López, D. Reliability Study of Diagnostic Tests for Functional Hallux Limitus. *Foot Ankle Int.* **2020**, *41*, 457–462. [CrossRef]
57. Nawoczenski, D.A.; Cook, T.M.; Saltzman, C.L. The effect of foot orthotics on three-dimensional kinematics of the leg and rearfoot during running. *J. Orthop. Sports Phys. Ther.* **1995**, *21*, 317–327. [CrossRef]
58. Kapandji, A.I. *Fisiología Articular T2: Miembro Inferior*; Médica Panamericana: Madrid, Spain, 2007.
59. Açak, M. The effects of individually designed insoles on pes planus treatment. *Sci. Rep.* **2020**, *10*, 1–6. [CrossRef] [PubMed]

5.3. ESTUDIO 3: Effect of foot orthoses on angular velocity of the foot of feet

sensors

MDPI

Article

Effect of Foot Orthoses on Angular Velocity of Feet

Juan Luis Florenciano Restoy [1], Jordi Solé-Casals [1,*] and Xantal Borràs-Boix [2,*]

[1] Data and Signal Processing Research Group, University of Vic—Central University of Catalonia, 08500 Vic, Spain
[2] Sport Exercise and Human Movement, University of Vic—Central University of Catalonia, 08500 Vic, Spain
* Correspondence: jordi.sole@uvic.cat (J.S.-C.); xantal.borras@uvic.cat (X.B.-B.)

Abstract: There is some uncertainty regarding how foot orthoses (FO) affect the biomechanics of the lower extremities during running in non-injured individuals. This study aims to describe the behavior of the angular velocity of the foot in the stride cycle measured with a low-sampling-rate IMU device commonly used by podiatrists. Specific objectives were to determine if there are differences in angular velocity between the right and left foot and to determine the effect of foot orthoses (FO) on the 3D angular velocity of the foot during running. The sample was composed of 40 male adults (age: 43.0 ± 13.8 years, weight: 72.0 ± 5.5 kg, and height: 175.5 ± 7.0 cm), who were healthy and without any locomotor system alterations at the time of the test. All subjects use FO on a regular basis. The results show that there are significant differences in the transverse plane between feet, with greater differences in the right foot. Significant differences between FO and non-FO conditions were observed in the frontal and transverse planes on the left foot and in the sagittal and transverse planes on the right foot. FO decreases the velocity of the foot in dorsi-plantar flexion and abduction and increases the velocity in inversion. The kinematic changes in foot velocity occur between 30% and 60% of the complete cycle, and the FO reduces the velocity in abduction and dorsi–plantar flexion and increases the velocity in inversion–eversion, which facilitates the transition to the oscillating leg and with it the displacement of the center of mass. Quantifying possible asymmetries and assessing the effect of foot orthoses may aid in improving running mechanics and preventing injuries in individuals.

Keywords: foot velocity; inertial sensor; foot orthoses; IMU; foot kinematics

Citation: Florenciano Restoy, J.L.; Solé-Casals, J.; Borràs-Boix, X. Effect of Foot Orthoses on Angular Velocity of Feet. *Sensors* **2023**, *23*, 8917. https://doi.org/10.3390/s23218917

Academic Editors: Giovanni Saggio and Michael E. Hahn

Received: 21 July 2023
Revised: 24 October 2023
Accepted: 31 October 2023
Published: 2 November 2023

1. Introduction

Inertial measurement units (IMU) are increasingly being used in the study of running, as they allow data capture in a variety of situations, such as indoors in the laboratory, in the clinic, or outdoors [1,2]. However, it is still more common to perform running analysis under laboratory and treadmill running conditions [1].

Numerous IMUs are available, with the most common ones allowing data capture along all three motion axes [1]. These IMUs offer a reliable means of monitoring spatiotemporal parameters accurately and precisely, facilitating various aspects of running analyses [1,3]. Previous studies have explored different aspects of running mechanics, including the running cycle, foot strike patterns [3–7], foot trajectory [8], or the angular displacement of the foot [9].

Advancements in IMU technology have made these devices smaller, lighter, and more affordable, rendering them suitable for sports motion analysis. High-sampling-rate IMUs have been used for research purposes [4,6,7,10–12], but podiatrists often work with IMUs that are affordable yet have sufficient functionality to assist in the final diagnosis; thus, they often use devices with low capture frequencies [5,8,9,13,14]. Studies have shown that a frequency of 12 Hz is sufficient to capture human body movement across different devices [15,16]. Some research even employed band-pass-filtered data with a 12 Hz cut-off frequency to analyze kinematics at varying running speeds (between 2.68 m/s and 4.47 m/s) [17]. Furthermore, satisfactory classification of human activity can be achieved with sampling frequencies as low as 15 Hz [14,18]. In sports and biomechanics, a recommended IMU sample rate is at least 20 Hz [19,20] with a 20 Hz cut-off frequency for lowpass filters observed to enhance prediction accuracy [10]. Florenciano et al. [9] have reported that a sampling rate of 30 Hz is sufficient to capture the foot movement when running at low speed, and the reported results are consistent with the other literature [3,21–23].

The position of the sensor depends on the anatomical area to be explored, with some studies placing the sensor on the tibia [24]. According to Benson et al. [1], this is the most common area for IMU placement. However, for the study of the foot, the sensor has usually been placed on the heel [3,4] and on the instep or upper part of the shoe [4,25,26]. Researchers have found the foot to be a suitable location for capturing spatiotemporal parameters and angular kinematic variables [26].

During running, the body tissues must withstand impact forces up to four times greater compared to walking, and the repetitive nature of steps, about 600 times per kilometer covered, puts individuals at risk for injury [27–29]. Structural and functional factors, such as the shape of the plantar arch and passive or supported range of motion in the longitudinal axis (IN-EV), are assessed during clinical podiatric examinations, and if imbalances or excessive ranges of motion are observed, insoles may be recommended for runners, even in the absence of injury, since they are believed to mitigate excessive plantar and joint loads imposed by ground reaction forces [28,30].

Foot orthoses (FO) are among the most widely used external supports for the treatment of musculoskeletal disorders [31]. FOs have shown promise in providing pain relief and increasing ankle stability [32,33]. At a kinematic level, it has been observed that they reduce maximal foot eversion [33] and mitigate hip adduction and knee internal rotation after foot strike in subjects with severe pronation [32]. They have also been observed to decrease tibial internal rotation in the transversal plane [34], as well as inversion–eversion (IN-EV) and abduction–adduction (ABD-ADD) ranges of motion, but not dorsi–plantar flexion (D-PF) [9]. As excessive supination during the stance period is considered a potential risk factor for injury among runners, reducing motion in the frontal plane, where supination occurs, can be viewed as a preventive measure to mitigate such risks [28,35]. Foot orthoses have the potential to benefit individuals without existing health issues by decreasing the range of motion in the foot's axes, all while not impeding the natural running technique [9].

FOs are actively utilized by athletes for both therapeutic and preventive purposes [9,32–34,36]. However, despite their widespread use, there is still some uncertainty regarding how FO affects the biomechanics of the lower extremities during running in non-injured individuals, and the mechanisms underlying this effect are not well understood. By understanding the behavior of the angular velocity of the foot during running and the effects of FO on it, as well as their potential benefits in reducing injury risk and improving performance during running, clinicians can make informed decisions on whether to recommend FO for healthy individuals, particularly those who engage in high-impact activities like running.

A healthy gait is assumed to be symmetrical, but asymmetries often exist [37]. Previous references have highlighted that having a limb asymmetry greater than 15% could be associated with an increased incidence of injury in both athlete and non-athlete populations [38]. Running shoes could significantly decrease the degree of rearfoot asymmetry [37], but only one study has been found on the effect of FOs on the asymmetry of running [9].

Finally, there are several studies that assess the effect of FOs on angular velocity, observing that angular velocity decreases when FOs are used [39,40]. These articles, however, do not use IMU as a data capture system. Thus, the purpose of this study is threefold: (1) to describe foot angular velocity in the running of healthy individuals in the no-FO condition, (2) to determine whether there are differences in angular velocity between the right and left feet with and without FO, and (3) to determine the effect of FO on foot angular velocity in each of the axes with a low-sampling-rate IMU device commonly used by podiatrists. It is hypothesized that there will be no difference in foot angular velocity when FO is used and that there will be no difference in angular velocity between feet. This article focuses on analyzing the effect of FOs from the perspective of non-injury and therefore from the point of view of prevention.

2. Materials and Methods

A total of 40 males (43.0 ± 13.8 years) without any locomotor injuries at the time of the test participated in the study. The physical characteristics of the subjects were recorded using a mechanical column scale with a height gauge (ADE, Hamburg, Germany) (175.5 ± 7.0 cm; 72.0 ± 5.5 kg). All of them had experience and accumulated mileage and were already familiar with the FOs for at least one year [9]. They were informed of the conditions of the study and signed an informed consent form prior to their participation. This study was approved by the local ethics review board (UVic-UCC, 09/2016) and followed the Declaration of Helsinki.

2.1. Procedure and Data Acquisition

The analysis procedure started with the placement of the sensors on the instep of the sports shoe, on each foot (Figure 1). This is a commonly used IMU in podiatry and physiotherapy with a triaxial accelerometer, gyroscope, and magnetometer (MotionPod 30 Hz, Grenoble, France) [41]. In addition to the Velcro® fastening system supplied, adhesive tape was used to better fix the device and reduce vibrations, as described in Florenciano et al. [9]. In both conditions tested, participants were wearing their usual running shoes, as it has been observed that changes in shoe hardness can affect lower limb kinematics [42,43].

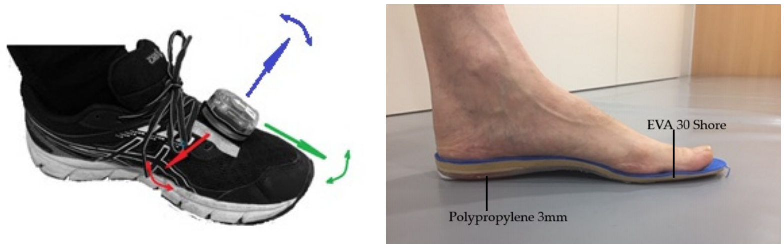


Figure 1. (**Left**): Position of the sensor attached to the instep of the running shoe. The triple orthogonal system represented by the arrows indicates the dorsi–planar flexion (red), abduction–adduction (blue), and eversion–inversion (green) movements. (**Right**): type of FOs used in both feet with the Polypropylene and the EVA layers.

Because the factory default settings were maintained in this study, and therefore the sampling rate was 30 Hz, the running speed was 2.5 m/s, near the transition between walking and running. Visual inspection ensured that none of the participants were walking instead of running. Raw data were filtered using the default factory settings as well. The same device and configuration were also used in [9], in which the amplitude of movement during the stance phase and the evaluation of the effects of FOs on foot kinematics were analyzed.

Subjects performed a 3 min warm-up run on a treadmill (BH Fitness G6414V SPORT, Alava, Spain) at 2.5 m/s (9 km/h) without any incline (measured with a spirit level). The warm-up also allowed the runners to familiarize themselves with the speed of the treadmill and the environment. Treadmill running is representative of running on the ground [44,45].

Once the warm-up period was completed, the participant rested for two minutes while the rest of the procedure was explained and the sensor was calibrated.

IMUs were calibrated according to the manufacturer's instructions. First, on the loading rail at the start of the session and, after being placed on the instep, adjusted to the relative reference system with the participant standing still and upright for 3 s. Thus, in a triple orthogonal relative reference system, the vertical axis captures the ABD-ADD movements of the foot, the longitudinal axis captures the IN-EV movements, and the D-PF movements are captured on the medial–lateral axis.

Two 20 s measurements were performed, the first one with participants wearing their usual running shoes and the second one with their usual running shoes and FO in both feet, at a speed of 2.5 m/s to meet IMU sampling rate requirements. The FOs used for each participant were those they used on a regular basis and with which they were already familiar. Following the IMU manufacturer's recommendations, data acquisition was performed for 20 s after the treadmill speed stabilized. This time is guaranteed using MotionPod to ensure data reliability and avoid measurement errors caused by data integration over time. The initial steps when passing between stationary and running phases were discarded, as were the deceleration steps at the end.

The manufacturing system of the FO (Figure 1) consists of a fast-setting wet plaster mold, adapted through a vacuum chamber [46]. Then, a paper pattern is made with the width and length of the foot that is adjusted to a Polypropylene (3 mm) sheet and heated in an oven at 180°. This allows the Polypropylene to be deformed and then adjusted to the plaster mold. The same process is performed with Ethylene–Vinyl Acetate (EVA) (6 mm thickness and 30 Shore hardness). EVA is heated up to 80º. EVA and Propylene shapes are bonded with approved glue. More details on the process can be found in the Supplementary Materials.

2.2. Data Processing and Analysis

The raw data provided using the IMU device, properly filtered with the acquisition device to meet the Nyquist requirements of the sampling rate, were processed with Matlab (Mathworks, Natick, MA, USA).

Using the sagittal plane movement (D-PF) as a reference, we determined the running stride, defined between the maximum and minimum D-PF points (Figure 2). These instants are taken as a reference to determine the stride in the frontal and transverse planes, as well as to extract the number of strides (N) of each subject, as conducted in Florenciano et al. [9,13].

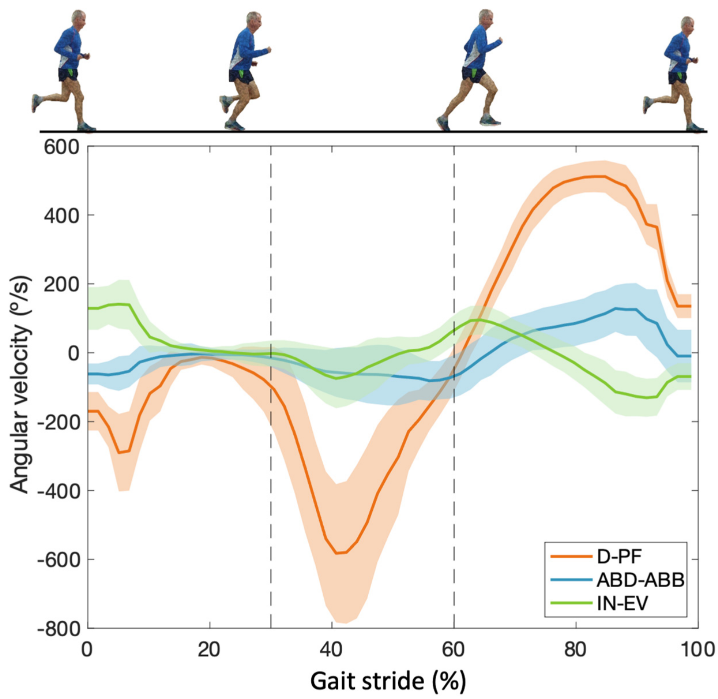


Figure 2. Velocity curves for dorsi–plantar flexion (D-PF), abduction–adduction (ABD-ADD), and inversion–eversion (IN-EV) movements. Negative values are plantar flexion, inversion, and abduction movements, while positive values are dorsiflexion, eversion, and adduction movements. Angular velocity increases when deviating from 0 and decreases when approaching 0.

The angular velocity on each axis of movement was determined by calculating the mean amplitude of each movement over time. This average amplitude was obtained by summing the movements during the stance phase of each step performed within the 20 s data collection period and dividing it by the total number of steps [9].

Mathematically, the data reduction process is described as follows:

1. Obtaining mean motion: Let $f(t)$, $a(t)$, and $p(t)$ be the functions representing the time evolution for the motions D-PF, ABD-ADD, and IN-EV, respectively. The mean motions are expressed as follows:

$$\overline{f}(t) = \frac{\sum_{i=1}^{N} f_i(t)}{N},\ \overline{a}(t) = \frac{\sum_{i=1}^{N} a_i(t)}{N},\ \overline{p}(t) = \frac{\sum_{i=1}^{N} p_i(t)}{N} \tag{1}$$

where the symbol—above the variable denotes mean value, and N is the number of total steps. Since each of the time evolution functions may have a slightly different number of points, linear interpolation was used to ensure that the number of points was the same for all of them before calculating the mean values given in Equation (1);

2. Obtaining the mean angular velocity: Let $\overline{wf}(t)$, $\overline{wa}(t)$, and $\overline{wp}(t)$ be the mean angular velocities of the D-PF, ABD-ADD, and IN-EV motions, respectively expressed in degrees/second. These velocities are obtained by deriving the original $f_i(t)$, $a_i(t)$, and $p_i(t)$, curves with respect to time. Then, the mean values are obtained by averaging them across steps:

$$\overline{wf}(t) = \frac{\sum_{i=1}^{N} \frac{\partial f_i(t)}{\partial t}}{N},\ \overline{wa}(t) = \frac{\sum_{i=1}^{N} \frac{\partial a_i(t)}{\partial t}}{N},\ \overline{wp}(t) = \frac{\sum_{i=1}^{N} \frac{\partial p_i(t)}{\partial t}}{N} \tag{2}$$

A first statistical analysis was conducted using open-source JASP 0.14 1.0 software, focusing on the mean angular velocity the impulse phase, which accounts for approximately 30% to 60% of the complete running cycle [47]. The 30% is the percentage of cycle where the foot ends the stationary phase and at 60% the stance phase ends [47]. Normality was tested using the Shapiro–Wilk test for the mean value of the subjects' data (angular velocity) [48]. A paired measures comparison (T-Student) was used. The confidence interval and significant alpha level were set at 95% and 0.05, respectively. To allow for better interpretation of the data, effect size (d-Cohen) was conducted using the following criteria [49]: insignificant—$d \leq 0.2$; small effect—$0.2 < d \leq 0.5$; medium effect—$0.5 < d \leq 0.8$; and large effect—$d > 0.8$.

The asymmetry between the left and right feet was calculated for the mean angular velocity on each axis of motion in both FO and non-FO conditions. To do so, Equation (3) was used. The minimum value (conversely, the maximum value) was chosen among the mean velocity values, considering both the left and right mean curves (for the specific axis motion). Therefore, the direction of the difference is not analyzed. This equation is accurate for the calculations of asymmetries in unilateral tests [50].

$$Asymmetry = -100\,\frac{minimum\ value}{maximum\ value} + 100 \tag{3}$$

A second statistical analysis was used considering the whole datapoints, from 0% to 100% of the gait stance. Statistical parametric mapping (SPM) was used to assess time series differences in angular velocity after the control period and after the custom-made FO intervention period. All SPM analyses were performed using the open-source code spm1d [19] (v.0.4, spm1d.org) in MATLAB (R2020B) (The Math-Works, Inc., Natick, MA, USA).

3. Results

3.1. Description of Velocity Curves

Figure 2 shows the angular velocity curves for D-PF, ABD-ADD, and IN-EV. The amplitude of angular velocity is greater for D-PF than for IN-EV and ABD-ADD. The minimum angular velocity (negative values) of plantar flexion, inversion, and abduction occurs between 10% and 30%. At 30%, the stationary phase of the foot ends and the heel rise begins [13]. This produces an increment in the angular velocity, with the maximum angular velocity observed between 30% and 40%.

The angular velocity of these movements decreases up to 60% of the cycle for plantar flexion and to 70% for inversion and abduction. From these points on, the movement moves to dorsal flexion, eversion, and adduction. With an increase in angular velocity (positive values) at the beginning and a decrease towards the end of the stroke cycle.

3.2. Differences between Feet

Table 1 presents the mean and standard deviation data for the angular velocity of each limb between 30% and 60% of the cycle. The left and right feet present similar values of angular velocity in D-PF; the right foot shows a lower velocity, although this difference is not statistically significant (p = 0.8 for the non-FO condition and p = 0.4 for the FO condition), and the effect size is practically insignificant (d = 0.027 for the non-FO condition and d = 0.122 for the FO condition). In the frontal plane (IN-EV) and transverse plane (ABD-ADD), the right foot exhibits a slightly higher velocity. These differences are not statistically significant for IN-EV (p = 0.4 for the non-FO condition and p = 0.9 for the FO condition), with an insignificant effect size (d = 0.129 for the non-FO condition and d = -8.14×10^{-4} for the FO condition), but they are significant for ABD-ADD (p = 0.009 for non-FO condition and p = 0.02 for FO). The effect size is small for both movements (d = 0.433 for the non-FO condition; d = 0.367 for the FO condition). These results confirm that in both D-PF and IN-EV movements, there is no significant difference in foot velocity for the non-FO and FO conditions; however, in ABD-ADD movements, the difference is significant in both the non-FO and FO conditions.

Table 1. Mean and standard deviation of the angular velocity for left and right legs in degree/second (°/s), between 30% and 60% of the cycle. Values of asymmetry as a percentage. * Indicates statistically significant differences ($p < 0.05$) between angular values when comparing the left and right legs.

Non-FO Condition	**Left**	**Right**	**Asymmetry**	***p*-Value**	**d-Cohen**
Dorsi–Plantar flexion	−324.3 ± 40.8	−323.4 ± 54.8	0.27 ± 25.5	0.800	0.027
Inversion–Eversion	−19.4 ± 30.5	−25.1 ± 42.6	22.7 ± 28.4	0.400	0.129
Abduction–Adduction	57.1 ± 44.2	−79.4 ± 42.8	28.0 ± 3.1	0.009 *	0.433
Orthoses (FO) Condition					
Dorsi–Plantar flexion	−319.1 ± 36.2	−315.3 ± 52.9	1.20 ± 31.5	0.400	0.122
Inversion–Eversion	−29.8 ± 34	−29.7 ± 44	0.13 ± 22.7	0.900	-8.14×10^{-4}
Abduction–Adduction	−49.4 ± 43.1	−68.4 ± 40.4	27.7 ± 6.2	0.020 *	0.367

3.3. Effect of the Foot Orthoses

Table 2 presents the mean and standard deviations of the angular velocity, showing the effect of the FO. One can observe that the FO decreases the angular velocity values in the D-PF movement, being statistically significant for the right foot (p = 0.01) but not for the left foot (p = 0.07). FO also decreases the angular velocity values in the ABD-ADD, with statistically significant results in both cases (p = 0.04 left foot, p = 0.02 right foot), while the effect size of the FO is small in both cases (d = 0.329 left foot, d = 0.385 right foot). The increase in velocity is statistically significant for IN-EV in the left foot (p = 0.006) with a small effect size (d = 0.455).

Figure 3 depicts (mean ± std) the effect of the FOs on the right and left feet and all the feet together in the whole data series. Differences are evident in several areas of the curve, although they are only statistically significant at specific points, according to the SPM.

Table 2. Mean and standard deviation of the angular velocity for each axis of motion in degree/second (°/s), between 30% and 60% of the cycle. * Indicates statistically significant difference ($p < 0.05$) when comparing non-FO and FO conditions. The arrows indicate whether the difference in angular velocity increases or decreases when comparing the conditions.

		Non-FO	FO	Difference	*p*-Value	d-Cohen
Left	Dorsi–Plantar flexion	-324.3 ± 40.8	-319.1 ± 36.2	5.2 ↓	0.070	0.286
	Inversion–Eversion	-19.4 ± 30.5	-29.8 ± 34	10.2 ↑	0.006 *	0.455
	Abduction–Adduction	57.1 ± 44.2	-49.4 ± 43.1	7.7 ↓	0.040 *	0.329
Right	Dorsi–Plantar flexion	-323.4 ± 54.8	-315.3 ± 52.9	8.1 ↓	0.010 *	0.405
	Inversion–Eversion	-25.1 ± 42.6	-29.7 ± 44	4.6 ↑	0.300	0.162
	Abduction–Adduction	-79.4 ± 42.8	-68.4 ± 40.4	11.0 ↓	0.020 *	0.385

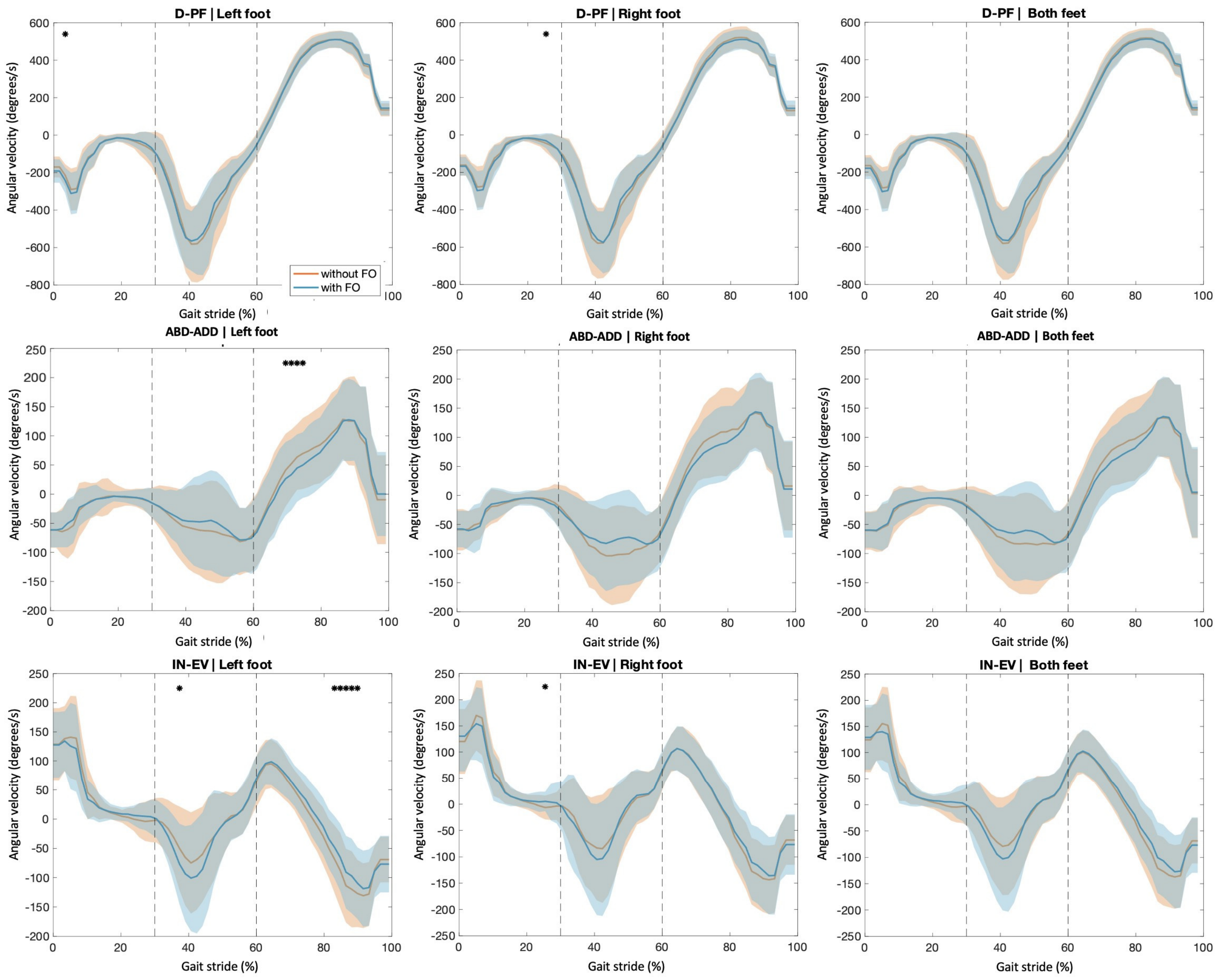


Figure 3. Angular velocity curves were plotted for each plane for the left foot, the right foot, and both feet together. Shaded areas indicate ±1 standard deviation from the mean value (solid line). The dashed lines delimit 30% and 60% of the cycle. * Indicates statistically significant difference ($p < 0.05$) when comparing non-FO and FO conditions for the whole datapoint using statistical parametric mapping.

4. Discussion

4.1. Description of the Velocity Curves

This study was able to describe the angular velocity curves on the three axes of motion (Figure 3). Other articles have described the angular displacement [13] but not the angular velocity. The amplitude of the angular velocity of the D-PF is greater than for IN-EV and ABD-ADD. As indicated by Florenciano et al. [13], angular displacement in the D-PF axis is notably greater, facilitating a more significant increase in velocity within this specific motion axis.

At a running velocity of 2.5 m/s, and from the angular velocity description, we suggest that around 30% of the cycle, the stationary phase of the foot ends and the heel rise begins. Plantar flexion velocity increases until 40% of the cycle, after which it decreases. At 60% of the cycle, the transition from plantar flexion to dorsiflexion occurs. This is according to Kapri et al. [47], who reported that the stance phase is up to 60% of the full cycle and the swing phase covers the final 40% of the total running cycle [51,52].

Fukuchi et al. [53] described that up to 50% of the full cycle of external rotation occurs in the pelvis and hip, and Kapandji [54] also described that, under normal conditions, the external rotation of the limb should coincide with the abducted position of the foot. The purpose of external rotation of the pelvis and hip, together with abduction, plantar flexion, and inversion of the foot, is to facilitate movement towards the opposite limb in a swing culminating in the aerial phase. From this kinematic perspective, and based solely on our experience and results, we suggest that it is in this percentage of the cycle that the center of mass (CoM) of the body shifts towards the other leg. During running, most of the forward force is generated by arm swings and leg swings [55].

4.2. Comparison between Feet

The results in Table 1 show the mean angular velocity between 30% and 60% of the cycle, which corresponds, in other words, to the impulse phase. The right foot presents higher angular velocity values than the left foot, both in IN-EV and ABD-ADD, being statistically significant in the ABD-ADD. Likewise, in a previous study, Florenciano et al. [9] found that the range of motion in the frontal and transverse planes was greater in the right foot than in the left.

Healthy gait is assumed to be symmetrical [37]. Because of this, the null hypothesis stated that there would be no difference between the feet in terms of the mean angular velocity; nevertheless, results indicate that the movement in the sagittal plane is symmetrical but not in the frontal and transverse planes when assessed with the non-FO and FO conditions.

Asymmetry between limbs has been observed in other studies; for example, Molitor [56] identified asymmetry in the range of motion in the medial arch and in the time between heel rise and toe off, while Gao et al. [57] observed asymmetry in medial foot pressure and the range of balance index between dominant and non-dominant limbs, with a preference for the dominant limb. The dominant limb was not assessed in the present study, so we cannot establish this type of relationship.

Previous references [38,58], have highlighted that having limb asymmetry greater than 15% is associated with a higher incidence of injury in both the athlete and non-athlete populations. Following this recommendation, our results show that IN-EV and ABD-ADD could be susceptible to injury. Interestingly, FOs do not influence the asymmetry of the velocity of movement, except for the IN-EV. This could be conditioned by the fact that FO was used on both limbs simultaneously. From these findings, it is suggested that the dynamic supporting functions of the right and left limbs should be considered.

4.3. Effect of FO Versus Non-FO Conditions

When considering mean angular velocity between 30% and 60% of the cycle, the results (Table 1) show that angular velocity in the D-PF and ABD-ADD planes is reduced with the use of FO, being significant in both the right and left feet. However, in the IN-EV, the angular velocity increases significantly between 30% and 60% of the cycle with the use of FO.

Other authors have also found that the use of customized FO has effects on the musculoskeletal structure, both in the modification of some biomechanical parameters and in the incidence of injury [36,59–61], For example, Betz et al. [59] observed a modification in ankle inversion with the use of FO. Bonanno et al. [36] found that FO reduced the incidence of tibial stress syndrome, patellofemoral pain, Achilles tendinopathy, and plantar fasciitis/plantar pain. Lucas-Cuevas et al. [61] observed a decrease in the overall stress experienced by the foot with the use of FO. Finally, Franklyn-Miller et al. [60] found that FO had an impact on injury reduction.

With respect to the angular velocity, these results are contradictory to Van Alsenoy et al. [40] and MacLean et al. [39], since they obtain a reduction in the angular velocity in the frontal plane (IN-EV). However, it must be considered that these studies use 3D videography for data capture.

The variations in angular velocity observed in this study may lead us to believe that the decrease in angular velocity in D-PF and ABD could slow down the external rotation of the extremity and increase the angular velocity of inversion at the beginning of the impulse phase. With these two conditions, the FO would be improving the transition to the opposite leg in the swing phase by achieving greater effectiveness in the displacement of the CoM. This could have a preventive effect on healthy subjects.

4.4. Limitations of the Study

The results and conclusions presented in this paper should be taken with caution. One of the limitations of this study relates to the sampling frequency used to collect the data. The sampling frequency used in this study has already been used in the past [3,9,13,21–23], but it inevitably limits the running speed of the subjects. Whether these results are valid for higher speeds, where the impact of the feet and the movement itself will be different, needs to be further investigated. In addition, the data were collected under controlled conditions (treadmill, in a flat position, and on a homogeneous surface), so other sampling frequencies would be needed in different situations. These results might not be directly extrapolated to outdoor running, especially in the case of cross-running, where the surface is very heterogeneous. Another limitation is the fact that the subjects in the study were healthy runners who used FOs on a regular basis. As a result, the effect of FOs on subjects who have never used them might be different. The limb dominance parameter was not assessed in the present study. From the data collected, we know that 87.5% of the participants had right foot dominance. However, due to the absence of homogeneous groups for comparison, we refrained from using this variable for the analysis. Therefore, we cannot establish any relationship between limb dominance and the results of foot asymmetry. Finally, another possible limitation would be the short data acquisition time. Although references have been found that use similar times [62,63], it could be a limitation to observe the effect of FO on angular foot velocity.

5. Conclusions

Based on the results of this study, we can conclude that (1) low-frequency IMUs allow obtaining foot angular velocity data that follow a certain pattern. This pattern has been described in the present study. (2) Kinematic variables exist between extremities during running in healthy subjects, and FOs did not mitigate the issue of asymmetry of the angular velocity, except for the IN-EV movement. It is suggested that the dynamic supporting functions of the right and left limbs should be considered. Furthermore, quantifying possible asymmetries could have implications for injury prevention during running. (3) The use of FO produces kinematic changes in foot angular velocity, mainly occurring between 30% and 60% of the complete cycle. The reduction in the angular velocity in ABD-ADD and D-PF and the increase in IN-EV could facilitate the displacement of CoM and the transition of the leg to the swing phase.

Supplementary Materials: The following supporting information can be downloaded at: https://www.mdpi.com/article/10.3390/s23218917/s1, Figure S1: Manufacturing process of the customised FO. References [64,65] are cited in the Supplementary Materials.

Author Contributions: Conceptualization, X.B.-B. and J.S.-C.; methodology, X.B.-B. and J.S.-C.; software, J.L.F.R.; formal analysis, J.L.F.R.; investigation, J.L.F.R. and X.B.-B.; data curation, J.L.F.R. and X.B.-B.; writing—original draft preparation, J.L.F.R.; writing—review and editing, X.B.-B. and J.S.-C.; supervision, X.B.-B. and J.S.-C.; project administration, X.B.-B. and J.S.-C. All authors have read and agreed to the published version of the manuscript.

Funding: This research received no external funding.

Institutional Review Board Statement: The study was conducted according to the guidelines of the Declaration of Helsinki and approved by the Institutional Ethics Committee of the University of Vic—Central University of Catalonia (UVic-UCC, 09/2016).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The datasets generated and analyzed during the current study are not publicly available due to ethics and privacy requirements but are available from the corresponding author on reasonable request.

Acknowledgments: This work was carried out as part of the doctoral program in Experimental Science and Technology at the University of Vic—Central University of Catalonia. The authors would like to acknowledge Juan Luis Florenciano Merino for his technical collaboration, and Pau Solé-Vilaró for the English proofreading of this manuscript.

Conflicts of Interest: The authors declare no conflict of interest.

References

1. Benson, L.C.; Räisänen, A.M.; Clermont, C.A.; Ferber, R. Is This the Real Life, or Is This Just Laboratory? A Scoping Review of IMU-Based Running Gait Analysis. *Sensors* **2022**, *22*, 1722. [CrossRef]
2. Ojeda, L.V.; Rebula, J.R.; Kuo, A.D.; Adamczyk, P.G. Influence of Contextual Task Constraints on Preferred Stride Parameters and Their Variabilities during Human Walking. *Med. Eng. Phys.* **2015**, *37*, 929–936. [CrossRef]
3. Boutaayamou, M.; Schwartz, C.; Stamatakis, J.; Denoël, V.; Maquet, D.; Forthomme, B.; Croisier, J.-L.; Macq, B.; Verly, J.G.; Garraux, G.; et al. Development and Validation of an Accelerometer-Based Method for Quantifying Gait Events. *Med. Eng. Phys.* **2015**, *37*, 226–232. [CrossRef]
4. Giandolini, M.; Poupard, T.; Gimenez, P.; Horvais, N.; Millet, G.Y.; Morin, J.-B.; Samozino, P. A Simple Field Method to Identify Foot Strike Pattern during Running. *J. Biomech.* **2014**, *47*, 1588–1593. [CrossRef]
5. Mariani, B.; Rouhani, H.; Crevoisier, X.; Aminian, K. Quantitative Estimation of Foot-Flat and Stance Phase of Gait Using Foot-Worn Inertial Sensors. *Gait Posture* **2013**, *37*, 229–234. [CrossRef]
6. Shiang, T.-Y.; Hsieh, T.-Y.; Lee, Y.-S.; Wu, C.-C.; Yu, M.-C.; Mei, C.-H.; Tai, I.-H. Determine the Foot Strike Pattern Using Inertial Sensors. *J. Sens.* **2016**, *2016*, 4759626. [CrossRef]
7. Sinclair, J.; Hobbs, S.J.; Protheroe, L.; Edmundson, C.J.; Greenhalgh, A. Determination of Gait Events Using an Externally Mounted Shank Accelerometer. *J. Appl. Biomech.* **2013**, *29*, 118–122. [CrossRef]
8. Suzuki, Y.; Hahn, M.E.; Enomoto, Y. Estimation of Foot Trajectory and Stride Length during Level Ground Running Using Foot-Mounted Inertial Measurement Units. *Sensors* **2022**, *22*, 7129. [CrossRef]
9. Florenciano Restoy, J.L.; Solé-Casals, J.; Borràs-Boix, X. IMU-Based Effects Assessment of the Use of Foot Orthoses in the Stance Phase during Running and Asymmetry between Extremities. *Sensors* **2021**, *21*, 3277. [CrossRef]
10. Alcantara, R.S.; Day, E.M.; Hahn, M.E.; Grabowski, A.M. Sacral Acceleration Can Predict Whole-Body Kinetics and Stride Kinematics across Running Speeds. *PeerJ* **2021**, *9*, e11199. [CrossRef]
11. Fasel, B.; Sporri, J.; Chardonnens, J.; Kroll, J.; Muller, E.; Aminian, K. Joint Inertial Sensor Orientation Drift Reduction for Highly Dynamic Movements. *IEEE J. Biomed. Health Inform.* **2018**, *22*, 77–86. [CrossRef]
12. Weygers, I.; Kok, M.; Konings, M.; Hallez, H.; De Vroey, H.; Claeys, K. Inertial Sensor-Based Lower Limb Joint Kinematics: A Methodological Systematic Review. *Sensors* **2020**, *20*, 673. [CrossRef]
13. Florenciano Restoy, J.L.; Solé-Casals, J.; Borràs-Boix, X. Descriptive Analysis of 3D Foot Motion Using Inertial Sensors. Comparison between Lower Extremities. *Rev. Española Podol.* **2021**, *32*, 1–5. [CrossRef]
14. Hegde, N.; Bries, M.; Sazonov, E. A Comparative Review of Footwear-Based Wearable Systems. *Electronics* **2016**, *5*, 48. [CrossRef]
15. Benoussaad, M.; Sijobert, B.; Mombaur, K.; Azevedo Coste, C. Robust Foot Clearance Estimation Based on the Integration of Foot-Mounted IMU Acceleration Data. *Sensors* **2015**, *16*, 12. [CrossRef]

16. Qiao, D.; Pang, G.K.H.; Kit, M.M.; Lam, D.C.C. A New PCB-Based Low-Cost Accelerometer for Human Motion Sensing. In Proceedings of the 2008 IEEE International Conference on Automation and Logistics, Qingdao, China, 1–3 September 2008; pp. 56–60.
17. Stiffler-Joachim, M.R.; Wille, C.M.; Kliethermes, S.A.; Johnston, W.; Heiderscheit, B.C. Foot Angle and Loading Rate during Running Demonstrate a Nonlinear Relationship. *Med. Sci. Sports Exerc.* **2019**, *51*, 2067–2072. [CrossRef]
18. Khan, A.; Hammerla, N.; Mellor, S.; Plötz, T. Optimising Sampling Rates for Accelerometer-Based Human Activity Recognition. *Pattern Recognit. Lett.* **2016**, *73*, 33–40. [CrossRef]
19. Sui, J.-D.; Chen, W.-H.; Shiang, T.-Y.; Chang, T.-S. Real-Time Wearable Gait Phase Segmentation for Running and Walking. In Proceedings of the 2020 IEEE International Symposium on Circuits and Systems (ISCAS), Seville, Spain, 12–14 October 2020; pp. 1–5.
20. Wagner, J.F. About Motion Measurement in Sports Based on Gyroscopes and Accelerometers—An Engineering Point of View. *Gyroscopy Navig.* **2018**, *9*, 1–18. [CrossRef]
21. Jones, H.S.R.; Stiles, V.H.; Verheul, J.; Moore, I.S. Angular Velocities and Linear Accelerations Derived from Inertial Measurement Units Can Be Used as Proxy Measures of Knee Variables Associated with ACL Injury. *Sensors* **2022**, *22*, 9286. [CrossRef]
22. Pagnon, D.; Domalain, M.; Reveret, L. Pose2Sim: An End-to-End Workflow for 3D Markerless Sports Kinematics—Part 2: Accuracy. *Sensors* **2022**, *22*, 2712. [CrossRef]
23. Sung, J.; Han, S.; Park, H.; Cho, H.-M.; Hwang, S.; Park, J.W.; Youn, I. Prediction of Lower Extremity Multi-Joint Angles during Overground Walking by Using a Single IMU with a Low Frequency Based on an LSTM Recurrent Neural Network. *Sensors* **2021**, *22*, 53. [CrossRef]
24. Wylde, M.J.; Callaway, A.J.; Williams, J.M.; Yap, J.; Leow, S.; Yong, L.C. Limb Specific Training Magnitude and Asymmetry Measurement to Discriminate between Athletes with and without Unilateral or Bilateral Lower Limb Injury History. *Phys. Ther. Sport* **2022**, *56*, 76–83. [CrossRef]
25. Elstub, L.J.; Nurse, C.A.; Grohowski, L.M.; Volgyesi, P.; Wolf, D.N.; Zelik, K.E. Tibial Bone Forces Can Be Monitored Using Shoe-Worn Wearable Sensors during Running. *J. Sports Sci.* **2022**, *40*, 1741–1749. [CrossRef]
26. Zrenner, M.; Küderle, A.; Roth, N.; Jensen, U.; Dümler, B.; Eskofier, B.M. Does the Position of Foot-Mounted IMU Sensors Influence the Accuracy of Spatio-Temporal Parameters in Endurance Running? *Sensors* **2020**, *20*, 5705. [CrossRef]
27. Lieberman, D.E.; Warrener, A.G.; Wang, J.; Castillo, E.R. Effects of Stride Frequency and Foot Position at Landing on Braking Force, Hip Torque, Impact Peak Force and the Metabolic Cost of Running in Humans Effects of Stride Frequency and Foot Position at Landing on Braking Force, Hip Torque, Impact Peak For. *J. Exp. Biol.* **2015**, *218*, 3406–3414. [CrossRef]
28. Fields, K.B.; Sykes, J.C.; Walker, K.M.; Jackson, J.C. Prevention of Running Injuries. *Curr. Sports Med. Rep.* **2010**, *9*, 176–182. [CrossRef]
29. Altman, A.R. A Kinematic Method for Footstrike Pattern Detection in Barefoot and Shod Runners Allison. *Gait Posture* **2021**, *35*, 298–300. [CrossRef]
30. Creaby, M.W.; May, K.; Bennell, K.L. Insole Effects on Impact Loading during Walking. *Ergonomics* **2011**, *54*, 665–671. [CrossRef]
31. Moisan, G.; Robb, K.; Mainville, C.; Blanchette, V. Effects of Foot Orthoses on the Biomechanics of the Lower Extremities in Adults with and without Musculoskeletal Disorders during Functional Tasks: A Systematic Review. *Clin. Biomech.* **2022**, *95*, 105641. [CrossRef]
32. Lack, S.; Barton, C.; Woledge, R.; Laupheimer, M.; Morrissey, D. The Immediate Effects of Foot Orthoses on Hip and Knee Kinematics and Muscle Activity during a Functional Step-up Task in Individuals with Patellofemoral Pain. *Clin. Biomech.* **2014**, *29*, 1056–1062. [CrossRef]
33. Mündermann, A.; Nigg, B.M.; Neil Humble, R.; Stefanyshyn, D.J. Foot Orthotics Affect Lower Extremity Kinematics and Kinetics during Running. *Clin. Biomech.* **2003**, *18*, 254–262. [CrossRef]
34. Nawoczenski, D.A.; Cook, T.M.; Saltzman, C.L. The Effect of Foot Orthotics on Three-Dimensional Kinematics of the Leg and Rearfoot During Running. *J. Orthop. Sports Phys. Ther.* **1995**, *21*, 317–327. [CrossRef]
35. Shih, Y.; Ho, C.-S.; Shiang, T.-Y. Measuring Kinematic Changes of the Foot Using a Gyro Sensor during Intense Running. *J. Sports Sci.* **2014**, *32*, 550–556. [CrossRef]
36. Bonanno, D.R.; Landorf, K.B.; Munteanu, S.E.; Murley, G.S.; Menz, H.B. Effectiveness of Foot Orthoses and Shock-Absorbing Insoles for the Prevention of Injury: A Systematic Review and Meta-Analysis. *Br. J. Sports Med.* **2017**, *51*, 86–96. [CrossRef]
37. Vagenas, G.; Hoshizaki, B. A Multivariable Analysis of Lower Extremity Kinematic Asymmetry in Running. *J. Appl. Biomech.* **1992**, *8*, 11–29. [CrossRef]
38. Impellizzeri, F.M.; Rampinini, E.; Maffiuletti, N.; Marcora, S.M. A Vertical Jump Force Test for Assessing Bilateral Strength Asymmetry in Athletes. *Med. Sci. Sports Exerc.* **2007**, *39*, 2044–2050. [CrossRef]
39. MacLean, C.; McClay Davis, I.; Hamill, J. Influence of a Custom Foot Orthotic Intervention on Lower Extremity Dynamics in Healthy Runners. *Clin. Biomech.* **2006**, *21*, 623–630. [CrossRef]
40. Van Alsenoy, K.; van der Linden, M.L.; Ryu, J.H.; Girard, O.; Al Raisi, L.; Santos, D. Isolated and Combined Effects of EVA and TPU Custom Foot Orthoses on Constant Speed, Treadmill Running Kinematics. *Front. Earth Sci.* **2023**, *11*, 1170661. [CrossRef]
41. Seifert, L.; L'Hermette, M.; Komar, J.; Orth, D.; Mell, F.; Merriaux, P.; Grenet, P.; Caritu, Y.; Hérault, R.; Dovgalecs, V.; et al. Pattern Recognition in Cyclic and Discrete Skills Performance from Inertial Measurement Units. *Procedia Eng.* **2014**, *72*, 196–201. [CrossRef]

42. Lussiana, T.; Fabre, N.; Hébert-Losier, K.; Mourot, L. Effect of Slope and Footwear on Running Economy and Kinematics. *Scand. J. Med. Sci. Sports* **2013**, *23*, 246–253. [CrossRef]
43. Fischer, K.M.; Willwacher, S.; Hamill, J.; Brüggemann, G.-P. Tibial Rotation in Running: Does Rearfoot Adduction Matter? *Gait Posture* **2017**, *51*, 188–193. [CrossRef]
44. Fellin, R.E.; Manal, K.; Davis, I.S. Comparison of Lower Extremity Kinematic Curves during Overground and Treadmill Running. *J. Appl. Biomech.* **2010**, *26*, 407–414. [CrossRef]
45. Firminger, C.R.; Vernillo, G.; Savoldelli, A.; Stefanyshyn, D.J.; Millet, G.Y.; Edwards, W.B. Joint Kinematics and Ground Reaction Forces in Overground versus Treadmill Graded Running. *Gait Posture* **2018**, *63*, 109–113. [CrossRef]
46. Jin, Y.; Plott, J.; Chen, R.; Wensman, J.; Shih, A. Additive Manufacturing of Custom Orthoses and Prostheses—A Review. *Procedia CIRP* **2015**, *36*, 199–204. [CrossRef]
47. Kapri, E.; Mehta, M.; Singh, K. Biomechanics of Running: An Overview on Gait Cycle. *Int. J. Phys. Educ. Fit. Sports* **2021**, *10*, 1–9. [CrossRef]
48. Hanusz, Z.; Tarasinska, J.; Zielinski, W. Shapiro–Wilk Test with Known Mean. *Revstat Stat. J.* **2016**, *14*, 89–100. [CrossRef]
49. Cohen, J. *Statistical Power Analysis for the Behavioral Sciences*; Routledge: Hillsdle, MN, USA, 2013; ISBN 9781134742707.
50. Bishop, C.; Turner, A.; Maloney, S.; Lake, J.; Loturco, I.; Bromley, T.; Read, P. Drop Jump Asymmetry Is Associated with Reduced Sprint and Change-of-Direction Speed Performance in Adult Female Soccer Players. *Sports* **2019**, *7*, 29. [CrossRef]
51. Buschbacher, R.M.; Prahlow, N.D.; Dave, S.J. *Sports Medicine and Rehabilitation: A Sport-Specific Approach*; Buschbacher, R.M., Prahlow, N.D., Dave, S.J., Eds.; Lippincott Williams & Wilkins: Philadelphia, PA, USA, 2008.
52. Dugan, S.A.; Bhat, K.P. Biomechanics and Analysis of Running Gait. *Phys. Med. Rehabil. Clin. N. Am.* **2005**, *16*, 603–621. [CrossRef]
53. Fukuchi, C.A.; Fukuchi, R.K.; Duarte, M. A Public Dataset of Overground and Treadmill Walking Kinematics and Kinetics in Healthy Individuals. *PeerJ* **2018**, *6*, e4640. [CrossRef]
54. Kapandji, A.I. *The Physiology of the Joints. Volume 2, The Lower Limb*; Churchill Livingstone: New York, NY, USA, 2012.
55. Mann, R.A.; Hagy, J. Biomechanics of Walking, Running, and Sprinting. *Am. J. Sports Med.* **1980**, *8*, 345–350. [CrossRef]
56. Molitor, S.L. Characterizing the Effect of Lower Limb Dominance on Foot Kinematics During Walking. Ph.D. Thesis, Vanderbilt University, Nashville, TN, USA, 2021.
57. Gao, Z.; Mei, Q.; Xiang, L.; Baker, J.S.; Fernandez, J.; Gu, Y. Effects of Limb Dominance on the Symmetrical Distribution of Plantar Loading during Walking and Running. *Proc. Inst. Mech. Eng. P J. Sport Eng. Technol.* **2022**, *236*, 17–23. [CrossRef]
58. Barber, S.D.; Noyes, F.R.; Mangine, R.E.; McCloskey, J.W.; Hartman, W. Quantitative Assessment of Functional Limitations in Normal and Anterior Cruciate Ligament-Deficient Knees. *Clin. Orthop. Relat. Res.* **1990**, *255*, 204–214.
59. Betz, J.; Klingspor, C.; Seel, T. IMU-Based Assessment of Ankle Inversion Kinematics and Orthosis Migration. In Proceedings of the 2019 41st Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC), Berlin, Germany, 23–27 July 2019; pp. 6395–6400.
60. Franklyn-Miller, A.; Wilson, C.; Bilzon, J.; McCrory, P. Foot Orthoses in the Prevention of Injury in Initial Military Training. *Am. J. Sports Med.* **2011**, *39*, 30–37. [CrossRef]
61. Lucas-Cuevas, A.G.; Pérez-Soriano, P.; Llana-Belloch, S.; Macián-Romero, C.; Sánchez-Zuriaga, D. Effect of Custom-Made and Prefabricated Insoles on Plantar Loading Parameters during Running with and without Fatigue. *J. Sports Sci.* **2014**, *32*, 1712–1721. [CrossRef]
62. Glassbrook, D.J.; Fuller, J.T.; Alderson, J.A.; Doyle, T.L.A. Foot Accelerations Are Larger than Tibia Accelerations during Sprinting When Measured with Inertial Measurement Units. *J. Sports Sci.* **2020**, *38*, 248–255. [CrossRef]
63. Johnson, C.D.; Outerleys, J.; Davis, I.S. Relationships between Tibial Acceleration and Ground Reaction Force Measures in the Medial-Lateral and Anterior-Posterior Planes. *J. Biomech.* **2021**, *117*, 110250. [CrossRef]
64. Florenciano Restoy, J.L. Adaptación de soportes plantares por ordenador. *Peu* **2002**, *22*, 150–153.
65. Florenciano Restoy, J.L. Tratamiento de la fascitis plantar mediante el sistema podocomputer. *Podol. Clín.* **2011**, *12*, 20–23.

CAPÍTULO 6

CAPÍTULO 6. DISCUSIÓN GENERAL

6.1. Descripción de las curvas

Descripción de las curvas de desplazamiento angular

En el Estudio-I se realiza una descripción del desplazamiento angular del pie en los tres planos del espacio. La obtención de las curvas de desplazamiento angular ha permitido determinar el movimiento normal del pie durante la carrera en personas sanas y no lesionadas. Las curvas nos permiten establecer patrones de movimiento típicos del pie.

En la figura 3 del Estudio-I se presentan la gráfica de desplazamiento angular del pie. En relación con esta figura podemos observar, coincidiendo con Kapri et al., (2021), que el 60% del ciclo es periodo de apoyo y el 40% periodo de oscilación. En el plano sagital el movimiento del pie es en flexión plantar hasta 60% del ciclo y en el 40% restante se mueve en flexión dorsal, en el plano frontal el pie se mueve en eversión hasta el 30% del ciclo y cambia a inversión hasta el 60%, se mueve en eversión hasta el 80% de ciclo y a inversión hasta el 100% del ciclo. En el plano horizontal el movimiento del pie es en abducción hasta el 60% del ciclo y mantiene la aducción hasta el 100% del ciclo.

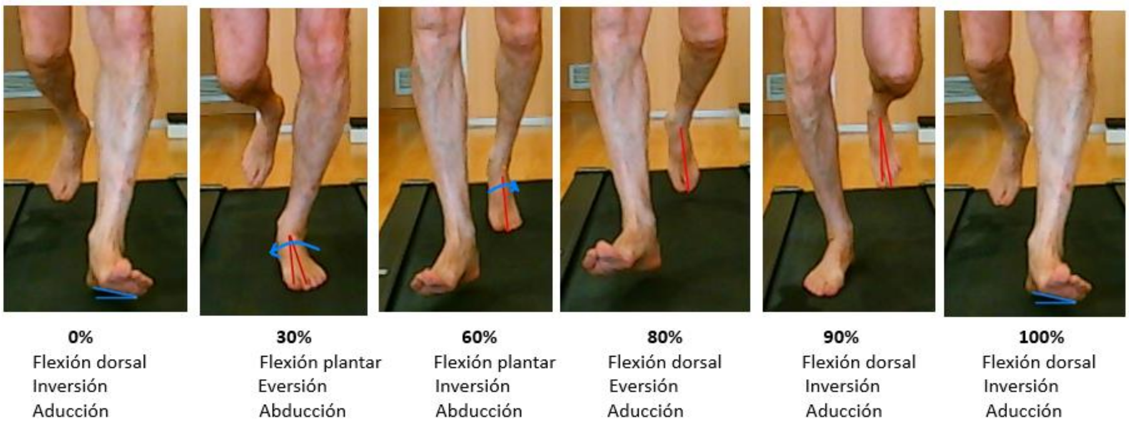


Figura 3. Movimientos del pie en % de ciclo completo en los planos sagital, frontal y transverso desde una visión frontal. Las líneas y curvas en color azul son los movimientos de eversión e inversión y las líneas rojas los movimientos de abducción y aducción.

Más al detalle, la Figura 3 muestra una secuencia fotográfica del movimiento del pie. Se observa que: (1) En el instante 0% del ciclo el pie está en flexión dorsal, inversión y aducción. (2) En el 30% del ciclo el pie presenta flexión plantar, eversión y abducción. (3) En el 60% del ciclo el pie presenta flexión plantar, inversión y aducción. (4) En el 80%

del ciclo el pie realiza flexión dorsal, eversión y abducción. (5) En el 90% flexión dorsal, inversión y aducción y (6) en el 100% vuelve a estar en flexión dorsal, inversión y aducción como al inicio del ciclo de carrera.

Descripción de las curvas en la velocidad angular

Como se detalla en el Estudio-III a partir del procesamiento de los datos obtenidos por el IMU también se pudo obtener la velocidad angular del pie para cada uno de los tres planos de movimiento. La gráfica de la velocidad angular del pie se puede observar en la Figura 2 del Estudio-III.

En esta gráfica, cuando la trayectoria se separa del 0 la velocidad angular aumenta y cuando se aproxima al 0 la velocidad angular disminuye. Los valores negativos indican flexión plantar, inversión y abducción, mientras que los valores positivos indican flexión dorsal, eversión y aducción

La velocidad angular del pie en flexión plantar es la que permite definir el resto de los ejes. Para ello, el proceso de calibración del sensor es determinante; al calibrar el sensor con el sujeto en bipedestación estática, se entiende que la velocidad del pie es 0 º/s.

Cuando en el plano sagital la velocidad del pie es 0 º/s se inicia la “fase estacionaria”. Cuando la curva se sitúa en al eje “X” indica que la planta del pie no se mueve y que es la extremidad en apoyo la que se desplaza hacia delante. Esto se evidencia en el 30% del ciclo, donde la velocidad angular hacia la flexión plantar aumenta al alejarse del 0 º/s. Entre el 30% y el 40% del ciclo se produce un evento muy interesante y es que el talón se separa del suelo indicando el inicio de la fase de impulso. A partir del 40% hasta el 60% del ciclo la velocidad angular en flexión plantar disminuye, siendo la pendiente de la curva proporcional a la aceleración angular del pie.

6.2. Movimientos del pie en 3D y su relación con la cadena cinemática de la extremidad inferior.

El movimiento del pie forma parte de la cinemática de la extremidad inferior y debe participar de forma efectiva en el desplazamiento del tronco. La biomecánica de la carrera se puede analizar mediante la anatomía de las extremidades superiores e inferiores (Van Oeveren et al., 2021).

Al correr, la parte superior del cuerpo (cabeza, brazos y parte superior del tronco) y la parte inferior del cuerpo (parte inferior del tronco y piernas) se mueven en direcciones contrarias en el eje longitudinal. La cintura escapular y la cintura pélvica se mueven en la misma dirección, pero en sentido contrario, generando momentos de inercia que facilita el desplazamiento del cuerpo hacia adelante (Hamner & Delp, 2013)

La fisiología articular de la articulación de la cadera es importante para entender el desplazamiento del cuerpo y por qué del movimiento 3D del pie. Los movimientos de extensión, rotación y abducción de la articulación de la cadera no tienen lugar alrededor del eje anatómico-diafisario del fémur, sino más bien del *eje mecánico* (Figura 4) que se proyecta desde el centro de la articulación de la cadera al centro de la articulación de la rodilla (Peterson et al., 2000). Esta disposición del eje mecánico del fémur facilita que, en el periodo de oscilación de la extremidad, el trocánter mayor del fémur se desplace hacia delante y hacia dentro, generando dos movimientos simultáneos (flexión, rotación interna de la cadera) (Peterson et al., 2000). Los músculos que se inserten por delante del eje mecánico actuarán como rotadores internos del fémur y, al contrario, en el periodo de apoyo el trocánter mayor del fémur se desplaza hacia atrás y hacia fuera generando igualmente dos movimientos simultáneos (extensión y rotación externa de la cadera), los músculos que inserten por detrás del eje mecánico actuarán como rotadores externos del fémur invirtiendo las acciones musculares (Kapandji, 2012).

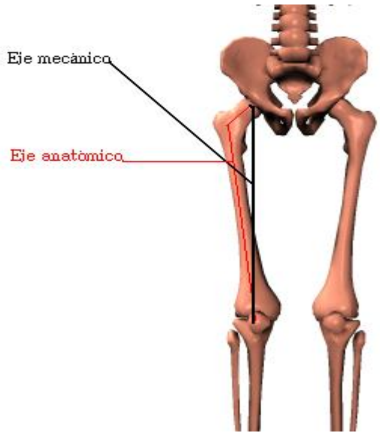


Figura 4. Ejes anatómico y mecánico del fémur.

Comparación movimiento 3D del pie y el movimiento 3D de la cadera

En la Figura 5 se observa el desplazamiento angular de la cadera en cada uno de los ejes de movimiento descritos por Fukuchi et al. (2018). Estos autores analizaron el movimiento de la cadera en 42 adultos sanos mediante videografía 3D (Fukuchi et al., 2018).

La extensión de cadera se produce entre el 0% y el 60% del ciclo coincidiendo con los valores obtenidos en el Estudio-I de flexión plantar del pie. Acto seguido y siguiendo la trayectoria ascendente de la curva se puede comprobar que la flexión de la cadera, según la gráfica, se acompaña de flexión dorsal del pie hasta completar el ciclo.

También el movimiento de eversión del pie coincide con el inicio de rotación interna de la cadera (trayectoria ascendente) en el 10% del ciclo, hasta la estabilización del pie en el 30%. El movimiento siguiente es en inversión hasta el 60% del ciclo, igual que hace la cadera mediante la rotación externa hasta el 60% del ciclo.

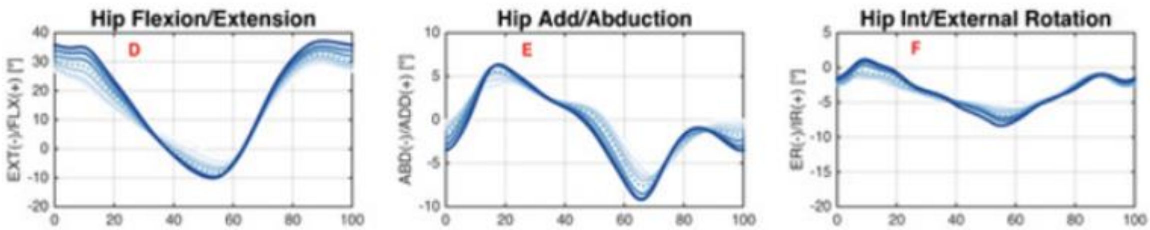


Figura 5. *Imagen superior, gráficas de la cadera en ciclo completo* (Fukuchi et al., 2018).

El movimiento de ABD del pie es hasta el 60% del ciclo, también se refleja en la cadera dado que también esta se desplaza en ABD hasta el 60% del ciclo. Es decir, en la medida que la cadera desplaza el centro de gravedad del cuerpo (CGC) hacia la otra pierna en abducción también el pie se mueve en abducción hasta el 60% del ciclo.

Esta relación cinemática entre el movimiento 3D del pie y el movimiento 3D de la cadera, sugiere, en gran medida, la eficacia de los movimientos del pie y de la cadera en el desplazamiento hacia delante del tronco (plano sagital) y en la alternancia entre extremidades planos, frontal y transverso.

Por lo tanto, en el periodo de apoyo tanto la cadera como el pie se extienden, la cadera rota externamente (inversión del pie) y la ABD de la cadera se complementa con la ABD del pie para desplazar el tronco hacia la extremidad en oscilación, cuya finalidad es facilitar el desplazamiento anterior del tronco y la alternancia en las extremidades minimizando el desplazamiento del CGC y con ello, tanto el gasto energético como el equilibrio físico de la persona. (Lee & Chou, 2006) determinaron los ángulos de inclinación en los planos sagital y frontal del CGC, encontraron que la basculación de la pelvis del plano frontal ABD-ADD de la pelvis y la cadera, es una variable independiente de la velocidad de marcha y el nivel de actividad.

6.3. Análisis de la asimetría

Asimetrías en desplazamiento angular

Los resultados en el Estudio-II permiten observar que en el plano sagital no existe diferencia entre pies en la amplitud de movimiento de desplazamiento angular, sin embargo, sí que se ve afectado el índice de simetría en los planos frontal y transverso puesto que el porcentaje de asimetría es de 31.9 ± 18.2 % para EV-IN y 21.7 ± 14.56.2 % para la ABD-ADD en pie derecho respectivamente.

Al comparar estos datos con los obtenidos de asimetría en el Estudio-I se constata que el porcentaje de asimetría para la amplitud de movimiento es menor en ciclo completo que en el periodo de apoyo, siendo el valor en la EV-IN del porcentaje de asimetría de 14.3 ± 4 % y el de ABD-ADD es de 15.2 ± 16.6 %, mayor en pie derecho respectivamente.

Estas diferencias podrían sustentarse en las investigaciones de Viel (2002). En sus estudios sobre el trabajo excéntrico y concéntrico de los músculos de la extremidad inferior durante el apoyo, proponen que los miembros inferiores no contactan con el suelo a la misma intensidad, habiendo un miembro inferior más amortiguador y otro más propulsor (Viel, 2002).

Kong & De Heer (2008) comprobaron en 6 atletas de élite keniatas a cinco velocidades de carrera (3.5 – 5.4 m/s) utilizando un sistema de captura de movimiento, el tiempo de toma de contacto con el suelo del pie derecho fue más breve (170 – 212 ms) en comparación con el pie izquierdo (177 – 220 ms). No se observaron diferencias bilaterales en las otras variables de movimiento o de fuerza.

Pulido-Valdeolivas et al. (2013) presentaron una muestra de 27 niños entre 5-13 años datos de normalidad de escolares españoles donde se observan asimetrías entre los lados izquierdo y derecho.

Asimetrías en la velocidad angular

En el Estudio-III los resultados mostraron que la velocidad del pie en el plano sagital era prácticamente la misma con diferencias en la asimetría insignificantes, por el contrario, los resultados si fueron significativos en la velocidad del pie en ABD con valores de $p=0.009$ con calzado y $p=0.02$ con OP, siendo en ambos casos las asimetrías importantes 28 ± 3.1 % con calzado y 27.7 ± 6.2 % con OP, en cambio, en el movimiento de IN-EV se detecta una singularidad, pues mientras la asimetría con calzado es de 22.7 ± 28 %, la asimetría con OP casi desaparece con un valor de 0.13 ± 22.7 %, lo que sugiere que la velocidad del pie en IN-EV se simetriza con OP.

Movimiento asimétrico y lesión.

Rodríguez Sánchez (2016) en una revisión bibliográfica concluye que, aun existiendo una relación entre la pronación anormal del pie y alteraciones musculoesqueléticas del miembro inferior no se ha encontrado evidencia científica suficiente para establecer una fuerte relación causa-efecto. Se necesita más investigación a fin de aumentar la evidencia científica en este campo.

En la Figura 6 se ha observado que el tobillo se mueve durante el apoyo primero en pronación y después en supinación, en el Estudio-II se confirma que la diferencia en los movimientos latero-mediales mayores en pie derecho tanto con calzado como con OP, están dentro de la normalidad, sugerimos que se debe considerar que al aumentar diferencias en los grados de movimiento entre pies sea motivo de lesión.

6.4. Efecto de las ortesis plantares en la cinemática del pie

Efecto de las ortesis en el desplazamiento angular

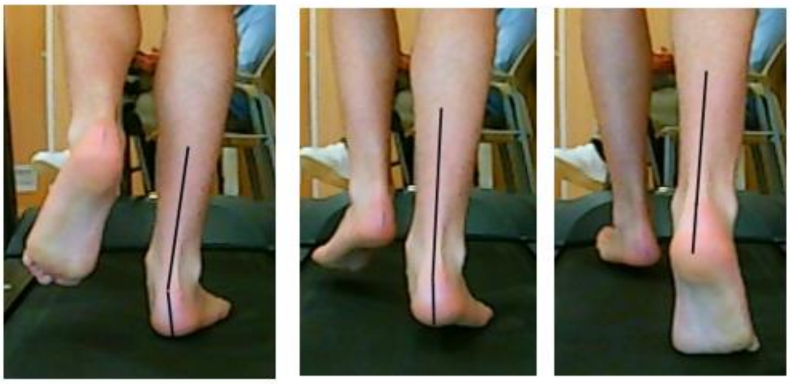
Los resultados de la amplitud de movimiento 3D en el Estudio-II constatan una tendencia de la OP a reducir la amplitud de movimiento en los planos frontal y transverso, pero no en el plano sagital.

Esta reducción de la amplitud de movimiento latero-mediales del pie se debe explicar desde la compleja articulación del tobillo, dado que, los movimientos en el pie no son puros sino combinados.

Desde el inicio de contacto en 0% del ciclo hasta la fase de apoyo medio, 30% del ciclo, el pie efectúa una evidente eversión y abducción (pronación del tobillo), sin embargo, entre el 30% y 60% del ciclo, el pie eleva el talón en inversión y aducción (supinación del tobillo) (Figura 6).

A la luz de los resultados obtenidos en los movimientos latero-mediales del Estudio-II se sugieren que la OP reduce la amplitud de movimiento de pronación y también la amplitud de movimiento de supinación.

Estos resultados son parecidos a los de Escamilla-Martínez et al. (2013) que observaron una tendencia al aumento de la pronación después de la carrera (3,3 m/s) de 60 min en 30 hombres sanos corredores habituales. Los sujetos ganaron 2 puntos en el índice de postura del pie y el apoyo total y las áreas de contacto del talón medial aumentaron, al igual que las presiones debajo de la cabeza del segundo metatarsiano y el talón medial.



30% 40% 60%

Figura 6. *Movimiento del tobillo en visión posterior. Primera imagen: 30% del ciclo fase de apoyo medio, máxima eversión del pie (pronación del tobillo). Segunda imagen: 40% del ciclo, en el inicio de impulso, inicio de inversión del pie (inicio de la supinación del tobillo). Tercera imagen: 60% del ciclo, final de la fase de impulso, máxima inversión del pie (finalización de la supinación del tobillo). Fuente propia.*

En el Estudio-II la reducción de la amplitud de movimiento está vinculada a los resultados de tiempo de apoyo y ciclo completo, donde se observa un ligero aumento del tiempo de contacto y de oscilación, y una discreta reducción en el número de zancadas. Se sugiere que la reducción en el número de zancadas podría disminuir el gasto energético.

Esta observación puede ayudarnos a comprender el funcionamiento del pie, prevenir lesiones y diseñar ortesis plantares efectivas en el deporte.

Relación de los materiales de la OP con los materiales del calzado deportivo.

Otra cuestión importante que se debe destacar son los materiales empleados en la fabricación de la ortesis plantar deportiva. Dinato et al. (2021) compararon el calzado minimalista con el calzado con entresuela de poliuretano más termoplástico y comprobaron que el calzado deportivo con poliuretano y termoplástico aumentaba la economía a baja velocidad de carrera. Sin embargo, la economía de carrera no mejoro a la velocidad promedio sostenida durante una contrarreloj de 3 km (∼15 km/h), lo que finalmente resultó en un rendimiento de carrera similar en comparación con las zapatillas minimalistas. Por lo tanto, podría sugerirse que la mejora de la economía de carrera observada con el material del calzado diseñado para mejorar el retorno de energía podría ser más relevante que el calzado minimalista para carreras de distancias más largas (≥5 km).

Como se ha comentado, en el Estudio-II hubo una tendencia a disminuir el número de zancadas con las OP deportivas. Estas OP son fabricadas con materiales como es el termoplástico (polipropileno) y con material de amortiguación como es el Vinil-Etil-Acetato (EVA), por lo que sugerimos un efecto similar de la OP deportiva al calzado deportivo con suela de poliuretano más termoplástico.

Efecto de la ortesis plantar en la velocidad angular

El Estudio-III tenía como objetivo determinar el efecto de la OP en la velocidad angular del pie en cada uno de los ejes de movimiento del pie durante la carrera.

En la figura 2 del Estudio-III se observa la velocidad angular del pie para los tres ejes de movimiento, tanto en el pie izquierdo como en el pie derecho, con un denominador común, el pie experimenta cambios en la velocidad que coincide con el movimiento de la otra extremidad en oscilación. Estos eventos se producen entre el 30% y el 60% del ciclo y más concretamente en el 40% del ciclo con una tendencia de la OP a reducir la velocidad en ambos pies en abducción y flexión plantar. Si tomamos en consideración los datos del Estudio-II también existe una tendencia a reducir los grados de amplitud. Sugerimos que esta circunstancia es debida al momento de inercia de la otra pierna en oscilación.

En cuanto al movimiento de IN-EV sucede más bien lo contrario, aumenta la velocidad en el pie con OP, esto se explica desde la propia naturaleza del periodo de apoyo, la rotación externa de la extremidad en la extensión de la cadera, la inversión el pie gira hacia fuera y facilita la rotación externa de la cadera. Por lo tanto, a la luz de estos resultados de los movimientos 3D de la cinemática del pie, se sugiere que el efecto de la OP modifica, influye y favorece el comportamiento de la cadena cinemática de la extremidad inferior.

6.5. Bibliografia

Dinato, R. C., Cruz, R., Azevedo, R. A., Hasegawa, J. S., Silva, R. G., Ribeiro, A. P., Lima-Silva, A. E., & Bertuzzi, R. (2021). Footwear designed to enhance energy return improves running economy compared to a minimalist footwear: Does it matter for running performance? Brazilian Journal of Medical and Biological Research, 54. https://www.scielo.br/j/bjmbr/a/NssmgYqXPdrKcvBY8s9zcjS/

Fukuchi, C. A., Fukuchi, R. K., & Duarte, M. (2018). A public dataset of overground and treadmill walking kinematics and kinetics in healthy individuals. PeerJ, 6, e4640.

Hamner, S. R., & Delp, S. L. (2013). Muscle contributions to fore-aft and vertical body mass center accelerations over a range of running speeds. Journal of biomechanics, 46(4), 780-787.

Kapandji, IA. (2012). Fisiología articular. Miembro inferior. 6a ed. Editorial Maloine; 2012. Pág—Buscar con Google (6a ed.). Editorial Maloine.

Kapri, E., Mehta, M., & Singh, K. (2021). Biomechanics of running: An overview on gait cycle. International Journal of Physical Education, Fitness and Sports, 10.

Kong, P. W., & De Heer, H. (2008). Características Antropométricas, de la Zancada y de la Fuerza en Corredores Keniatas de Fondo. Grupo sobrenetrenamiento.com. https://g-se.com/caracteristicas-antropometricas-de-la-zancada-y-de-la-fuerza-en-corredores-keniatas-de-fondo-1194-sa-v57cfb271d4aa3

Lee, H.-J., & Chou, L.-S. (2006). Detection of gait instability using the center of mass and center of pressure inclination angles. Archives of physical medicine and rehabilitation, 87(4), 569-575.

Peterson, F., Kendall, E., & Geise, P. (2000). Músculos. Pruebas Funcionales. Postura y Dolor (4a ed.). Marban. https://marbanlibros.com/products/kendall-s-musculos-pruebas-funcionales-postura-y-dolor

Pulido-Valdeolivas, I., Gómez-Andrés, D., Martín-Gonzalo, J. A., López-López, J., Gómez-Barrena, E., Hernández, J. S., & Rausell, E. (2013). Parámetros de marcha en una muestra de referencia de escolares sanos españoles: Descripción multivariante y asimetrías entre ciclos izquierdos y derechos. Neurología, 28(3), 145-152.

Rodríguez Sánchez, D. (2016). Consecuencias de la pronación anormal del pie en alteraciones del miembro inferior: Una revisión bibliográfica. https://ruc.udc.es/dspace/handle/2183/18515

Van Oeveren, B. T., De Ruiter, C. J., Beek, P. J., & Van Dieën, J. H. (2021). The biomechanics of running and running styles: A synthesis. Sports Biomechanics, 1-39. https://doi.org/10.1080/14763141.2021.1873411

Viel, E. (2002). La marcha Humana, la Carrera y el Salto. Biomecánica, exploraciones, normas y alteraciones. Editorial Masson. Barcelona, 2.

CAPÍTULO 7

CAPÍTULO 7. CONCLUSIONES GENERALES

7.1. Conclusiones por objetivos generales

Estas conclusiones están relacionadas con los objetivos generales de la tesis doctoral:

Conclusión 1. Se han obtenido las medias tanto de amplitud de movimiento como de velocidad angular en ciclo completo y en el periodo de apoyo.

Conclusión 2. Se ha constatado que existen asimetrías en extremidades en las variables cinemáticas con y sin ortesis plantar.

Conclusión 3. El efecto cinemático de la ortesis plantar durante el periodo de apoyo es, tanto en la amplitud de movimiento, como en la velocidad angular.

7.2. Conclusiones por estudios

Estudio-I: Existen diferencias con más grados de amplitud de movimiento en pie derecho que en pie izquierdo en los planos frontal y transverso. En el plano sagital no hay diferencias.

Estudio 2: Se observa una tendencia de la OP a reducir los grados de amplitud en los movimientos de los planos frontal y transverso con mayor tiempo de apoyo y ciclo completo con OP. Con disminución en el número de zancadas con OP.

Estudio 3: La velocidad del pie en apoyo se ve afectada a partir de la fase estacionaria por la inercia de la pierna oscilante, entre el 30% y 60% del ciclo. La OP modifica la cinemática disminuyendo la velocidad del pie en los planos sagital y transverso y aumentando la velocidad del pie en el plano frontal, mejorando la transición hacia la pierna oscilante y con ello tanto el gasto energético como el equilibrio dinámico.

Los resultados obtenidos en esta tesis doctoral sugieren que, la utilización de la IMU ha sido determinante para poder comprender en el periodo de apoyo y oscilación el movimiento 3D del pie.

CAPÍTULO 8

CAPÍTULO 8. LIMITACIONES Y LÍNEAS DE FUTURO

8.1. Aportación del doctorado

Una de las contribuciones más importantes de este doctorando ha consistido en implementar una metodología con la que se han podido obtener de cada participante las medias de movimiento 3D del pie. Todos los sujetos corrieron en una cinta de correr, a la misma velocidad 2,5 m/s y durante 20 s, por lo que el número de zancadas fue similar para todos ellos. Una vez caracterizado este movimiento en un participante, se siguió la misma metodología para el resto de los participantes, tanto en ciclo completo (apoyo y oscilación) como solo en el periodo de apoyo. Se pudo obtener una visión panorámica, tanto de la amplitud de desplazamiento angular, como de la velocidad angular del pie.

A nuestro modo de ver esta visión es particularmente original puesto que es un primer paso para el desarrollo de nuevas investigaciones que permitan establecer los grados de normalidad en el pie en el ámbito de la carrera.

Otra contribución, no menos importante de esta metodología, ha sido verificar el efecto cinemático de la OP en la mecánica del pie, por lo que se sugiere que esta metodología se podría aplicar a otras IMU, al calzado deportivo y también, poder obtener en tiempo real el efecto de la OP en el corredor.

8.2. Limitaciones de la investigación

Debemos ser cautos con los resultados y conclusiones presentados en este doctorado. Una limitación se debe relacionar con la frecuencia de muestreo utilizada para recolectar los datos. La frecuencia de muestreo utilizada en este estudio es baja, debido a las capacidades técnicas del sensor, pero ya se ha utilizado en el pasado (Boutaayamou et al., 2015; Jones et al., 2022; Pagnon et al., 2022; Sung et al., 2021). También se debe tener en cuenta como limitación que Bioval solo obtiene datos del movimiento lineal, es decir, no podemos hacer giros ni a derecha ni a izquierda.

La velocidad de carrera de 2,5 m/s de los sujetos es una velocidad de carrera lenta. Es necesario investigar si estos resultados son válidos para velocidades de carrera más altas.

También hay que mencionar que estos resultados no se extrapolen directamente a las carreras al aire libre, especialmente en el caso de las carreras de montaña, donde la superficie es muy heterogénea.

Otra limitación es el hecho de que los sujetos del estudio eran corredores sanos que usaban OP de forma regular. Como resultado, el efecto de los OP en sujetos que nunca los han usado podría ser diferente.

El parámetro de dominancia de extremidades no fue evaluado en el presente estudio. A partir de los datos recogidos, sabemos que el 87,5% de los participantes tenían dominancia del pie derecho. Sin embargo, debido a la ausencia de grupos homogéneos para la comparación, nos abstuvimos de utilizar esta variable para el análisis. Por lo tanto, no podemos establecer ninguna relación entre la dominancia de las extremidades y los resultados de la asimetría del pie. Por último, otra posible limitación sería el corto tiempo de adquisición de datos. Aunque se han encontrado referencias que utilizan tiempos similares (Glassbrook et al., 2020; Johnson et al., 2021), podría ser una limitación observar el efecto de la OP sobre la velocidad angular del pie.

8.3. Líneas de futuro

Una investigación interesante sería, manteniendo los mismos objetivos que en el presente doctorado, determinar la cinemática del pie durante la marcha a 1,25 m/s. Con lo que se podría obtener patrones de normalidad interesantes desde el punto de vista biomecánico y si el efecto cinemático de la OP es similar al de la carrera a 2,5 m/s.

Una posible línea de investigación sería poder determinar si se obtienen los mismos resultados solo en sujetos zurdos. Además, esta investigación podría desencadenar otras cuestiones interesantes, por ejemplo, en el caso que los zurdos mantuvieran los valores obtenidos en el Estudio-I y el pie derecho tuviera más grados de amplitud de movimiento que en el pie izquierdo, se deberían encontrar causas diferentes a la lateralidad.

Otra línea de investigación sería comprobar si las mujeres también presentan diferencias entre el pie derecho y el izquierdo observadas en el Estudio-I. Así como comprobar si la mujer puede beneficiarse de la ortesis de pie, dado que, las mujeres suelen presentar diferencias en la anatomía en comparación con los hombres. Al evaluar el efecto de la ortesis de pie exclusivamente en mujeres corredoras, se podría obtener una visión más precisa y específica de los beneficios potenciales que estos dispositivos podrían ofrecer a esta población.

8.4. Bibliografía

Boutaayamou, M., Schwartz, C., Stamatakis, J., Denoël, V., Maquet, D., Forthomme, B., Croisier, J.-L., Macq, B., Verly, J. G., & Garraux, G. (2015). Development and validation of an accelerometer-based method for quantifying gait events. Medical engineering & physics, 37(2), 226-232.

Glassbrook, D. J., Fuller, J. T., Alderson, J. A., & Doyle, T. L. A. (2020). Foot accelerations are larger than tibia accelerations during sprinting when measured with inertial measurement units. Journal of Sports Sciences, 38(3), 248-255. https://doi.org/10.1080/02640414.2019.1692997

Johnson, C. D., Outerleys, J., & Davis, I. S. (2021). Relationships between tibial acceleration and ground reaction force measures in the medial-lateral and anterior-posterior planes. Journal of biomechanics, 117, 110250.

Jones, H. S., Stiles, V. H., Verheul, J., & Moore, I. S. (2022). Angular velocities and linear accelerations derived from inertial measurement units can be used as proxy measures of knee variables associated with ACL injury. Sensors, 22(23), 9286.

Pagnon, D., Domalain, M., & Reveret, L. (2022). Pose2Sim: An end-to-end workflow for 3D markerless sports kinematics—part 2: accuracy. Sensors, 22(7), 2712.

Sung, J., Han, S., Park, H., Cho, H.-M., Hwang, S., Park, J. W., & Youn, I. (2021). Prediction of lower extremity multi-joint angles during overground walking by using a single IMU with a low frequency based on an LSTM recurrent neural network. Sensors, 22(1), 53.