ALESSANDRA BENTO MATIAS

Efeitos de um treinamento da musculatura do pé sobre os aspectos biomecânicos da corrida: um ensaio clínico randomizado

Tese apresentada à Faculdade de Medicina da Universidade de São Paulo para obtenção do título de Doutor em Ciências

Programa de Ciências da Reabilitação
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Ao meu avô Antônio Bento e a minha mãe Regina.

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“Não te deixes destruir…
Ajuntando novas pedras
e construindo novos poemas.
Recria tua vida, sempre, sempre.
Remove pedras e planta roseiras e faz doces. Recomeça.
Faz de tua vida mesquinha
um poema.
E viverás no coração dos jovens
e na memória das gerações que hão de vir.
Esta fonte é para uso de todos os sedentos.
Toma a tua parte.
Vem a estas páginas
e não entraves seu uso
aos que têm sede.”
(Cora Coralina, *Aninha e as pedras)*

NORMATIZAÇÃO ADOTADA

Esta tese está de acordo com as seguintes normas, em vigor no momento desta publicação:

Referências: adaptado de *International Committee of Medical Journals Editors* (Vancouver).

Universidade de São Paulo. Faculdade de Medicina. Divisão de Biblioteca e Documentação. *Guia de apresentação de dissertações, teses e monografias*. Elaborado por Anneliese Carneiro da Cunha, Maria Julia de A. L. Freddi, Maria F. Crestana, Marinalva de Souza Aragão, Suely Campos Cardoso, Valéria Vilhena. 3a ed. São Paulo: Divisão de Biblioteca e Documentação; 2011.

Abreviaturas dos títulos dos periódicos de acordo com *List of Journals Indexed in Index Medicus*.

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A corrida de rua é uma das atividades físicas mais populares, entretanto, as lesões relacionadas à corrida são altamente prevalentes e podem levar à interrupção da prática. A etiologia das lesões é multifatorial e cargas mecânicas e aspectos da biomecânica de membros inferiores parecem estar associadas a estas lesões. O pé é o primeiro segmento a entrar em contato com o solo e qualquer alteração na sua estrutura, função ou forma de aterrissar é capaz de alterar a mecânica do restante do membro inferior. Dependendo do tipo de aterrisagem na corrida (retropé, antepé e mediopé), o arco longitudinal absorve e armazena, diferentemente, as cargas recebidas como energia, contando com uma estrutura complexa que é mantida pela musculatura intrínseca e extrínseca, por ligamentos e pela aponeurose plantar. Dado o papel crucial do pé na corrida, esta tese teve como objetivo investigar aspectos biomecânicos das articulações do pé na corrida, bem como avaliar, por meio de um ensaio clínico randomizado e controlado, a eficácia de uma intervenção fisioterapêutica inovadora de 8 semanas, baseada em exercícios para fortalecimento da musculatura dos pés e tornozelos, na cinemática do pé e nas forças e taxas de impacto durante a corrida, em 87 corredores fundistas recreacionais. Antes desse ensaio clínico principal, desenvolvemos estudos complementares para melhor compreender a biomecânica do pé em diferentes aterrissagens da corrida e investigar a usabilidade e confiabilidade do modelo multisegmentar do pé. A primeira etapa na construção desta tese foi a elaboração de um programa de exercícios para os pés e a concepção e design do protocolo do ensaio clínico randomizado e controlado. A segunda etapa foi avaliar a confiabilidade, usabilidade e acurácia das medidas do modelo multisegmentar do pé na corrida que são um importante desfecho do ensaio clínico. Concluímos que a repetibilidade interexaminadores do modelo é menor na corrida em relação ao andar. Propusemos e testamos uma nova configuração de marcas do modelo do pé para a avaliação do arco longitudinal medial que apresentou uma confiabilidade menor em relação à configuração original, porém, a variabilidade de todos os ângulos, quando projetados tridimensionalmente, sempre foi menor do que quando projetados bidimensionalmente. Também investigamos a correlação e a acurácia das variações do modelo multisegmentar do pé e da nova proposição de marcas para a avaliação do arco longitudinal medial com medidas radiográficas clínicas padrões. Constatamos que a nova proposição utilizando a tuberosidade do navicular como vértice do arco é a medida que fornece a estimativa mais acurada do arco, quando comparada às medidas radiográficas. Na terceira etapa desta tese, investigou-se como o tipo de aterrissagem na corrida (antepé ou retropé) influenciaria a biomecânica do pé. Verificou-se que a forma como o pé entra em contato com o solo na corrida determina diretamente como será o comportamento cinemático do restante dos segmentos do pé na fase de apoio. Ainda nessa etapa, observamos, surpresos, que o primeiro pico da força vertical e as taxas de carga em alguns corredores de antepé assemelharam-se às de corredores de retropé. Na última etapa desta tese, concluímos que a intervenção fisioterapêutica de 8 semanas para o pé foi eficaz para modificar os padrões cinemáticos das articulações do tornozelo, tarso-metatársica, médio-társica e metatarso-falangeana do hálux, bem como de alguns fatores de risco biomecânicos para lesões, tais como: o movimento do arco longitudinal e o ângulo do retropé. No entanto, não houve efeito nas forças de impacto e na taxa de carga durante a corrida. As mudanças observadas na cinemática das articulações do pé podem ser responsáveis pela redução na incidência de lesões relacionadas à corrida observadas após o programa de treinamento dos pés em corredores recreacionais.

Descritores: Corrida; Pé; Biomecânica; Terapia por exercício; Pesquisa de reabilitação; Articulações do pé.

ABSTRACT

Matias, AB. *Effects of foot core strengthening on running foot biomechanics: a randomized controlled trial* [thesis]. São Paulo: "Faculdade de Medicina, Universidade de São Paulo"; 2021.

Running is one of the most popular physical activities, however, running-related injuries are highly prevalent and can lead to discontinuation of practice. The etiology of the injuries is multifactorial and mechanical loads and biomechanical aspects of the lower limbs seem to be associated with these injuries. The foot is the first segment to interact with the ground and any change in its structure, function or landing can alter the mechanics of the remainder of the lower limb. Depending on the type of the footstrike pattern (rearfoot, forefoot and midfoot), the longitudinal arch absorbs and stores the loads received as energy differently, relying on a complex structure that is maintained by the intrinsic and extrinsic muscles, ligaments, and plantar aponeurosis. Given the crucial role of the foot in running, this thesis aimed to explore the biomechanical aspects of the foot during running and, to investigate, through a randomized controlled clinical trial, the effectiveness of an innovative foot-core strengthening program of 8 weeks on foot-ankle kinematics and impact forces during running in recreational long-distance runners. Prior to this clinical trial, we developed further studies to better understand the biomechanics of the foot at different running footstrike patterns and assessed the usability and reliability of the multisegment foot model. The first step in the construction of this thesis was the development of a foot exercises program and the conception and design of the randomized controlled clinical trial protocol. The second step of this thesis was to assess the reliability, usability, and accuracy of the measurements of the multisegment foot model during running, which are an important outcome of the clinical trial. We conclude that the inter-examiner repeatability of the foot model is lower in running than in walking. We proposed and tested a new configuration of the skin marker-based multi-segment foot model for the evaluation of the medial longitudinal arch (MLA). We found that this new proposition had a lower reliability compared to the original configuration, but the variability of all angles with 3D projections was always smaller than the variability of 2D projections. We also assessed the correlation and accuracy of the variations and of the new proposition of the skin-marker based measures of MLA deformation with respect to standard clinical radiographic measures, used as reference. We found that the new proposition using the navicular tuberosity as the MLA vertex provided the most accurate estimate of the MLA when compared to radiographic measurements. The third step of this thesis investigated how the type of footstrike pattern (forefoot or rearfoot) would influence the foot biomechanics during running. It was found that the way the foot interact with the ground determines the kinematic behavior of the rest of the foot segments during stance phase. We were surprised to find that the first peak of the vertical force and the load rates in some forefoot runners were similar to those in rearfoot runners. In the last step of this thesis, we concluded that the physical therapy intervention was effective in modifying the kinematic patterns of the ankle, tarso-metarsal, midtarsal and metatarso-phalangeal joints; as well as some biomechanical risk factors for running injuries, such as the MLA movement and the rearfoot angle; but there was no effect on running impact forces and load rate. The observed changes in foot joint kinematics may be responsible for the reduction in running-related injuries incidence following the foot-core training program in recreational runners.

Descriptors: Running; Foot; Biomechanics; Exercise therapy; Rehabilitation research; Foot joints.

ESTRUTURA DA TESE

Esta tese de doutorado contém sete estudos originais (5 publicados e 2 em preparação para submissão), precedidos de uma introdução (capítulo I) que contém uma contextualização geral e uma discussão geral sobre os achados de todos os estudos ao final (capítulo VI). Todos os estudos estão relacionados à investigação da eficácia de uma intervenção terapêutica para os pés na biomecânica da corrida, em particular na cinemática do pé e na cinética da corrida de corredores fundistas. Investigamos como uma intervenção fisioterapêutica inovadora, focada na musculatura dos pés, modificaria a cinemática do pé e as forças e taxas de carga durante a corrida. Nossa hipótese é de que essa intervenção chamada “bottom-up”, por focar na extremidade distal do membro inferior, poderia alterar a mecânica da corrida. Porém, antes desse ensaio clínico principal, desenvolvemos estudos complementares para melhor compreender a biomecânica do pé na corrida e durante diferentes tipos de aterrissagem, e investigar a usabilidade, confiabilidade e acurácia do modelo multisegmentar do pé, que é o desfecho principal do ensaio.

O capítulo II apresenta o nosso primeiro artigo descrevendo o protocolo do ensaio clínico desenvolvido para avaliar a eficácia dessa intervenção terapêutica para os pés. Este artigo foi publicado na revista BMC Musculoskeletal Disorders (IF JCR=2.36).

O capítulo III mostra nossa preocupação e jornada científica para avaliar e aprimorar a confiabilidade, a usabilidade e a acurácia das medidas do modelo de pé multisegmentar que utilizaremos como desfecho do ensaio clínico e que foi desenvolvido no Instituto Ortopédico Rizzoli (Bolonha, Itália). Nesse instituto, realizei um estágio sanduíche (International Travel Grant Program from International Society of Biomechanics, 2016) com o Professor Dr. Alberto Leardini, que desenvolveu esse modelo multisegmentar do pé Rizzoli, e com o Dr. Paolo Caravaggi. Esse modelo é amplamente usado na literatura, mas ainda não possuía sua confiabilidade testada na corrida. Destarte, esse capítulo é dedicado ao aprimoramento da usabilidade, confiabilidade e acurácia do modelo de pé Rizzoli. Esse capítulo inclui 3 artigos: (1) apresenta o teste de confiabilidade do modelo de pé Rizzoli durante a corrida; (2) apresenta uma proposição e teste de uma nova configuração de marcadores para a avaliação de um novo modelo de arco longitudinal medial, e (3) apresenta uma avaliação da acurácia da nova configuração de marcas para a avaliação do arco com medidas radiográficas padrão. Em relação a este último artigo, as coletas de dados da corrida foram realizadas no Laboratório de Biomecânica e Postura Humana (LaBiMPH) no Brasil e as medidas de radiografia no Instituto Ortopédico Rizzoli. Esses 3 estudos são resultados de uma cooperação internacional de muito êxito e as autorias dos artigos foram compartilhadas, sendo que os 3 estudos foram publicados no Journal of Biomechanics (IF JCR=2.71).

O capítulo IV inclui 2 estudos originais que tratam a questão de como o tipo de aterrissagem (footstrike) na corrida (antepé ou retropé) influencia os padrões cinemáticos das articulações do pé e cinéticos da corrida. O primeiro estudo investigou e comparou os diferentes padrões de aterrissagem quanto à cinemática dos segmentos de pé. Este estudo foi realizado em colaboração com os pesquisadores Dr. Alberto Leardini e Dr. Paolo Caravaggi, do Instituto Ortopédico

Rizolli. O referido estudo foi publicado na revista Applied Sciences (IF JCR =2.69). O segundo estudo investigou as razões biomecânicas que fazem com que as forças verticais e as taxas de carga de alguns corredores de antepé sejam similares às de corredores de retropé. Este estudo foi desenvolvido durante o estágio de pesquisa no exterior, por 12 meses (FAPESP BEPE processo nº 2017/26844-1), realizado no Spaulding Running Center, Harvard University, Estados Unidos, sob supervisão da Professora Dra. Irene Davis. Com este estudo, ganhei o prêmio ISB-Sponsored Motor Control Group Student Award durante o Congresso Americano de Medicina Esportiva em 2019 (ACSM). O artigo desse estudo está em preparação para submissão e revisão dos coautores, mas já consta desta tese.

O capítulo V inclui o artigo referente aos desfechos secundários do ensaio clínico desenvolvido e mostra os resultados em relação à cinemática do pé e forças verticais e taxas de carga em corredores que receberam a intervenção, comparando-os com corredores que receberam uma intervenção placebo. Este artigo será submetido para a revista Scientific Reports (IF JCR = 4.379).

O VI e último capítulo apresenta uma discussão geral sobre os achados de todos os artigos, além das implicações clínicas, perspectivas futuras e as principais conclusões da tese. De maneira geral, os resultados do estudo sugerem que: (1) o modelo multisegmentar de pé Rizzoli é confiável para ser utilizado na corrida, apesar de menos confiável em relação ao seu uso para investigação do andar; (2) o modelo biomecânico de arco longitudinal que apresentou melhor reprodutibilidade é o original quando projetado tridimensionalmente; (3) porém, a nova configuração de marcas do modelo desenvolvida por nós, que usa a tuberosidade do navicular como vértice do arco, foi a mais acurada quando comparada às medidas radiográficas e às configurações originais. Finalmente, esta tese demonstrou que: (1) não é simplesmente a forma de aterrissagem na corrida que determina o padrão do impacto em corredores de antepé, mas a altura do calcanhar em relação ao solo no contato inicial, o tempo de chegada do calcanhar no solo e a sua aceleração; (2) a intervenção fisioterapêutica proposta para os pés modifica positivamente os padrões cinemáticos do pé e os fatores de risco biomecânicos para lesões, tais como: o movimento do arco longitudinal e o ângulo do retropé, embora não tenha tido efeito nas forças de impacto e na taxa de carga durante a corrida.

CAPÍTULO I -

CONTEXTUALIZAÇÃO

1.1 Corrida como estratégia democrática de atividade física

A corrida de rua tornou-se uma das atividades físicas mais populares nos últimos anos, provavelmente devido ao seu baixo custo e fácil acesso. A prática da corrida é motivada não apenas pela competição, bem-estar social e psíquico, mas, também, pela saúde física (1,2). No Brasil, estima-se que existam aproximadamente 10 milhões de corredores (3). Alguns dos benefícios da corrida de rua para a saúde incluem baixo risco de obesidade, hipertensão, dislipidemia, acidentes cerebrais, osteoartrite e até mesmo alguns tipos de câncer (2,4).

Entretanto, apesar de tantos benefícios, a corrida de rua apresenta uma alta prevalência de lesões entre os corredores de longa-distância. Estudos epidemiológicos relatam que as taxas de lesão em corredores estão entre 19 e 79% (5). Outros estudos reportam 6,8 a 59 lesões a cada 1000 horas de exposição à corrida (6–8). A ocorrência de lesões relacionadas à corrida é a principal causa de cessamento da participação neste tipo de atividade física e pode ocasionar aumento dos gastos com medicamentos e aumento das faltas no trabalho (9,10). A etiologia das lesões relacionadas à corrida é multifatorial (5) e esses fatores, possivelmente, interagem entre si (11). Entender os mecanismos que levam o corredor a ter estas lesões é fundamental para o desenvolvimento de estratégias de prevenção a aquelas relacionadas à corrida (12).

Os membros inferiores, especialmente o joelho (7-50%) e a perna (9-32%), são os locais mais afetados pelas lesões relacionadas à corrida (5). De acordo com van der Worp et al.(10), tecidos que apresentam uma perfusão pobre, como ligamentos, tendões e cartilagem, apresentam um risco maior de lesão quando ocorre um aumento da carga mecânica porque a adaptação destes tecidos é mais lenta quando comparada aos músculos. Hreljac et al.(13) sugeriram que as lesões poderiam ser evitadas não pela minimização do estresse aplicado `a estrutura biológica, mas pela otimização da carga e frequência desse estresse, ou seja, por ajustes na técnica da corrida e seu treinamento.

O pé é o primeiro segmento da cadeia cinética do membro inferior a entrar em contato com o solo e qualquer alteração na sua estrutura é capaz de causar um ajuste funcional, primeiro no tornozelo e depois na coordenação entre os segmentos adjacentes superiores (14). Em geral, o padrão de pisada da corrida é classificado em 3 grupos de acordo com a região do pé que faz o primeiro contato com o chão: (1) aterrissagem com o calcanhar (corredores de retropé), (2) aterrissagem com o antepé (corredores de antepé), e (3) aterrissagem com contato simultâneo do calcanhar e do antepé (corredores de mediopé) (15,16). Considerando apenas corredores que correm calçados, aproximadamente 75-92% são corredores habituais de retropé, entre 3,4-23,7% são corredores de mediopé e 1,4-1,8% são corredores de antepé (17–19).

O padrão de pisada é importante porque ele define a mecânica das extremidades inferiores ao aterrissar e seu progresso durante a fase de apoio, quando as forças externas irão agir no restante do corpo (20), definindo, por exemplo, um dos fatores de risco para lesões em corrida. A cinemática de tornozelo e pé ou os ângulos formados pelos marcadores são, geralmente, usadas para a classificação da pisada na corrida (21). Os corredores de retropé aterrissam com o tornozelo mais fletido, e o pé aterrissa à frente do centro de massa do corpo. Já os corredores de antepé aterrissam com o tornozelo em extensão e o pé aterrissa mais próximo ao centro de massa do corpo. Essas diferenças na posição e na orientação do pé produzem um padrão de força reação do solo distintos na primeira fase do apoio (22), embora a cinemática do tornozelo e pé durante a aterrissagem pareça interferir distintamente nas taxas de carga e foi o que investigamos na terceira etapa desta tese.

As altas taxas de carga de corredores de retropé estão presentes no transiente de impacto da força reação do solo vertical, sendo uma força de colisão abrupta de aproximadamente 1,5 a 3 vezes o peso corporal dentro dos primeiros 50ms da fase de apoio (23). Essas altas taxas de carga, tipicamente encontradas em corredores de retropé, têm sido associadas a lesões musculoesqueléticas e processos degenerativos (24,25). Corredores de mediopé ainda apresentam o pico de impacto (26,27). Corredores de antepé calçados são caracterizados geralmente pela ausência ou pela minimização do primeiro pico da forca reação do solo vertical (15,28–30). Os resultados obtidos na terceira etapa desta tese surpreenderam-nos, pois corredores de antepé apresentaram um padrão de impacto (força vertical e taxas de carga) semelhante a corredores de retropé, e as causas desse padrão não estiveram simplesmente relacionadas à forma como estes corredores aterrissavam na corrida, mas a como estava a altura do calcanhar em relação ao solo no contato inicial, qual era o tempo de chegada do calcanhar ao solo e qual era a aceleração do toque do calcanhar no contato inicial.

1.2 O papel do pé na corrida

Como o pé é a única parte do corpo humano que interage com o solo nas habituais habilidades de locomoção, sua estrutura provavelmente é uma das mais importantes durante a corrida em termos de capacidade de sustentação, absorção de impactos e distribuição de forças para o restante dos segmentos do corpo. O complexo pé-tornozelo é uma estrutura complexa capaz de ser ao mesmo tempo móvel e adaptável à cargas e, depois, é capaz de ser se tornar uma alavanca rígida durante a propulsão da corrida (31). Esse complexo é responsável pela interação entre o membro inferior e o solo, e o pé contribui para a distribuição adequada de cargas durante atividades em cadeia fechada, como a marcha humana (32). Durante a aterrisagem, a rigidez da perna é extremamente sensível a mudanças na rigidez do tornozelo que é a primeira a ser modulada. Uma das possíveis explicações para esse fenômeno é que a rigidez do tornozelo é menor do que a rigidez do joelho ou quadril (33). E, em um sistema com múltiplas molas como o corpo humano, o tornozelo, que é a mola menos rígida, é o que sofre o maior deslocamento em resposta à força aplicada e, assim, teria maior influência na rigidez total da perna (14).

O pé é formado por 26 ossos, 108 ligamentos, quatro camadas de musculatura intrínseca funcionando como estabilizadores locais (34), mais de 30 articulações com múltiplos graus de liberdade. A primeira camada muscular é formada pelo abdutor do hálux, flexor curto do dedo mínimo e abdutor do dedo mínimo. A segunda camada inclui o músculo quadrado plantar e os lumbricais. A terceira camada consiste nos músculos adutor do hálux, flexor curto do hálux e flexor curto dos dedos. Os músculos interósseos compõem a quarta camada.

A musculatura intrínseca faz parte do subsistema ativo da estabilização do pé proposto por Jam et al.(35) e Mckeon et al.(36), no qual o racional aplicado em relação à estabilização do complexo lombo-pélvico poderia ser usado em relação ao entendimento da estabilização do complexo do pé. Nesse conceito, o sistema de estabilização do pé é dividido em três subsistemas. O primeiro subsistema é composto pelas estruturas ativas e consiste nos músculos e tendões conectados ao pé. A musculatura extrínseca tem origem na perna, passa pelo tornozelo, insere-se no pé e age como um atuador global de movimento por meio de seus longos tendões que modulam a ação do subsistema passivo. O tendão calcâneo, por exemplo, modula a tensão da aponeurose plantar por meio da sua conexão em comum no calcâneo; assim, quando a tensão do músculo tríceps surae aumenta, a tensão da aponeurose plantar também aumenta (37).

O segundo subsistema do complexo do pé é o passivo e é composto por ossos, ligamentos e cápsulas articulares que mantêm os vários arcos do pé. Em uma configuração funcional, o pé é composto por 4 arcos distintos que incluem os arcos longitudinais medial e lateral e os arcos transversos anterior e posterior. As musculaturas intrínseca e extrínseca, juntamente com ligamentos e a aponeurose plantar, são responsáveis pela arquitetura do arco longitudinal medial, fazendo com que o pé humano seja único.

O terceiro subsistema é o neural e é formado pelos receptores sensoriais presentes na fáscia plantar, ligamentos, cápsulas articulares, músculos e tendões envolvidos nos subsistemas ativo e passivo. A estrutura e inserção da musculatura intrínseca têm desvantagem biomecânica para a produção de grandes e fortes movimentos articulares. No entanto, essa mesma estrutura e alinhamento muscular trazem uma vantagem de posicionamento para fornecer informação sensorial imediata sobre as mudanças articulares dos arcos do pé, por meio de mudanças no comprimento dos músculos intrínsecos, que facilitariam as respostas dos fusos neuromusculares.

Kelly et al.(38) observaram que a ativação da musculatura intrínseca leva ao aumento da resistência à deformação do arco quando o pé está envolvido em atividades dinâmicas com recebimento de carga. Existe uma interação de movimentos ocorrendo nas pequenas articulações do pé que permite que o arco longitudinal medial alongue-se e diminua sua altura durante a primeira metade da fase de apoio, absorvendo as cargas como forma de energia elástica pela compressão dos tecidos. Esse mecanismo permite que a energia mecânica seja temporariamente armazenada com a extensão dos ligamentos, músculos e tendões que envolvem o arco longitudinal medial. A energia mecânica é retornada no final da fase de apoio quando a resultante da força reação do solo diminui e as estruturas elásticas que estavam alongadas voltam a encurtar, permitindo que o arco longitudinal medial recolha-se. O recolhimento elástico da aponeurose plantar é passivo e contribui para a geração de trabalho positivo para a propulsão, conhecido como mecanismo de molinete (windlass mechanism), que, efetivamente, enrijece o arco longitudinal medial durante a extensão dos dedos (39,40).

O papel do pé durante a aterrisagem na corrida é de grande interesse da literatura devido a sua associação com a ocorrência de lesões relacionadas à corrida. Alguns estudos sugerem que o padrão de pisada de antepé pode contribuir para a economia da corrida (27) e para a prevenção de lesões relacionadas à corrida (41,42). Na aterrissagem de retropé, o corredor entra em contato com o chão com a região do calcanhar primeiro. Já na aterrisagem de antepé, o corredor inicia o contato com o chão com as cabeças do metatarso(23). Corredores habituais de antepé apresentam uma ativação muscular em torno do tornozelo diferente de corredores de antepé (23). A estabilização do tornozelo durante a aterrisagem é crucial para a atenuação de carga transmitida à articulação (43) e o padrão de pisada influencia diretamente nessa estabilização. A cinemática segmentar do pé, observando o padrão de pisada habitual de corredores durante a corrida, ainda é raramente reportada na literatura.

1.3 Métodos biomecânicos para avaliar o pé

A avaliação quantitativa mais precisa dos movimentos do pé é de extrema importância para o diagnóstico de doenças (44–47), para avaliação de calçados e órteses (48–51) e para a evolução de performance em gestos esportivos, como a corrida (52–54). O pé, quando modelado como um único segmento rígido, limita o conhecimento das contribuições e das interações dos seus múltiplos elementos na ocorrência de lesões.

Com o objetivo de avaliar os segmentos do pé, diversos métodos foram desenvolvidos na última década (55). A videofluoroscopia e os parafusos intraósseos são as medidas mais acuradas, porém estes dois métodos são extremamente invasivos (56). Sensores inerciais e escaneamento dinâmico 3D sem marcas são menos invasivos, mas a acurácia anatômica é comprometida (57). O uso de protocolos de marcas de pé multisegmentares para avaliação cinemática tridimensional tem sido cada vez mais comum para observar alterações nas articulações do pé e tornozelo, em populações saudáveis e pessoas com disfunções musculoesqueléticas (58).

Entre esses protocolos de marcas, o modelo multisegmentar de pé do Rizzoli destaca-se por ter a capacidade de rastrear o médio-pé e descrever o movimento do arco longitudinal medial com maior eficácia, sendo que seu uso tem sido aplicado a diversas populações e tarefas (56,59–61). Powell et al.(60) compararam os modelos multisegmentares de pé de Oxford e do Rizzoli durante a corrida e concluiram que o modelo do pé Rizzoli apresenta melhor desempenho durante a corrida por reportar diretamente o movimento do mediopé. O uso do modelo de pé Rizzoli em estudos sobre a biomecânica da corrida é extenso (60,62–68), porém a repetibilidade do modelo durante essa tarefa ainda não havia sido testada, e foi o que propusemos na etapa 2 desta tese. Como o pé desempenha um papel crucial durante a corrida, é essencial que se investigue os efeitos dessa atividade de alta energia na repetibilidade do modelo de marcas do pé do Rizzoli. Ainda, a avaliação dinâmica do arco longitudinal medial durante a corrida também é de extrema importância devido ao seu papel de atuação e transmissão de carga durante a corrida, o que ainda não havia sido testado até esta tese (39,69,70).

1.4 Intervenções terapêuticas para prevenção de lesões na corrida

Segundo a abordagem predominante em reabilitação, músculos mais fortes promovendo melhor controle postural, em especial a musculatura de quadril, ajudariam a reduzir a incidência de lesões em corredores. Ferber et al. (71), em 2010, observaram que a fraqueza e/ou a assimetria de força dos músculos abdutores e rotadores laterais de quadril aumentaram o risco de lesão em corredores, provavelmente em função de uma estabilização inadequada de quadril durante a corrida. Assim, essa abordagem terapêutica denominada "*top down*" sugere que o fortalecimento das musculaturas de quadril e do *core* ajudariam a reduzir os movimentos não sagitais nas articulações mais proximais que estariam mais associadas ao risco de lesão (72–74). Apesar dessa abordagem ter sido amplamente disseminada, ainda não há evidências contundentes de que sua prática altere a incidência de lesões relacionadas à corrida nas últimas décadas (34). Por outro lado, uma outra abordagem terapêutica denominada *"bottom-up",* cujo foco é o complexo articular do tornozelo-pé, poderia influenciar positivamente a biomecânica das articulações distais, e, por consequência, as mais proximais de joelho e quadril, potencializando a redução do risco de lesões (75–77).

Evidências na literatura mostram que intervenções simples podem ser aplicadas no pé, beneficiando até mesmo pessoas jovens e saudáveis. Kim et al.(78), em 2015, estudaram os efeitos de diferentes abordagens em indivíduos jovens com hálux valgo, com idades entre 19 e 29 anos. Os participantes foram divididos em 2 grupos: um grupo recebeu apenas órteses e o outro além das órteses, recebeu um treinamento que incluía um exercício de abdução dos dedos durante 20 minutos por dia, sendo realizado 4 dias por semana. O estudo mostrou que o grupo que recebeu o treinamento apresentou um aumento da área de secção transversa do abdutor do hálux e reduziu, significantemente, o ângulo de valgismo do hálux.

Alguns estudos com foco em exercícios para o fortalecimento da musculatura do pé observaram resultados positivos clínicos e de performance. Mulligan e Cook(79), em 2015, conduziram um estudo com pessoas saudáveis que executaram um exercício de fortalecimento da musculatura intrínseca do pé durante 4 semanas (short foot) e observaram uma redução significativa da queda do navicular, além de melhora na performance em tarefas de equilíbrio. Outro estudo avaliou o deslocamento do centro de pressão durante tarefas de equilíbrio e também observou melhora da performance após 4 semanas de exercícios focados no fortalecimento da musculatura intrínseca do pé (80).

Apesar desses achados positivos com o treinamento da musculatura intrínseca do pé, a eficácia desses exercícios para a redução de lesões em corredores e para modificação da biomecânica do pé na corrida não foi ainda testada. Uma musculatura intrínseca mais forte seria capaz de alterar a mecânica do pé na corrida, além de, potencialmente, prevenir lesões.

É esperado que uma melhora na morfologia e na força da musculatura do pé gere um aumento na capacidade do pé de atenuar mais impactos e de se comportar como uma alavanca mais rígida quando há aumento de carga, permitindo que o arco longitudinal medial armazene energia elástica durante a corrida de longa distância e a restitua na fase de propulsão, tornando até mesmo a corrida mais eficiente (81). Os efeitos de uma musculatura de pé mais forte poderiam ser vistos na melhora na absorção de impactos e na transmissão de forças entre o chão e o pé durante a corrida. Kelly et al.(38) mostraram que a ativação dos músculos abdutores do hálux, flexor curto dos dedos e quadrado plantar é maior na fase final do apoio no andar e na corrida. Essa maior ativação facilitaria a transmissão de forças do solo ao pé, permitindo que uma alta quantidade de força reação do solo seja transmitida em um curto período melhorando a funcionalidade do complexo articular na corrida. Nossa hipótese é de que músculos dos pés mais fortes alterariam a cinemática do pé de modo a minimizar o estresse aplicado às articulações de membros inferiores e, desse modo, diminuiria o número de lesões em uma população de corredores recreacionais de longa distância

Assim, na terceira etapa dessa tese buscou-se investigar os efeitos de uma intervenção terapêutica para os pés na incidência de lesões relacionadas à corrida, na biomecânica do pé e no comportamento de fatores de risco biomecânicos, tais como o ângulo do retropé e taxas de carga e forças verticais reação do solo, durante a corrida.

1.5 Objetivos

Objetivo geral

Dado o papel crucial do pé e seu potencial para melhorar sua funcionalidade durante a corrida, essa tese teve como objetivo geral investigar aspectos biomecânicos do pé na corrida, bem como investigar, por meio de um ensaio clínico randomizado e controlado, a eficácia de uma intervenção fisioterapêutica inovadora, baseada em exercícios para fortalecimento da musculatura dos pés e tornozelos, na cinemática do pé e nas forças e taxas de carga durante a corrida em corredores fundistas recreacionais.

Objetivos específicos

(i) Investigar a confiabilidade e a usabilidade das medidas do modelo multisegmentar do pé Rizzoli durante a corrida (Capítulo III, Estudo 3.1).

(ii) Propor e testar uma nova configuração de marcadores no modelo multisegmentar do pé para a avaliação do arco longitudinal medial (Capítulo III, Estudo 3.2).

(iii) Investigar a correlação e a acurácia dos modelos biomecânicos do pé originais e da nova proposição de marcas para a avaliação do arco longitudinal medial com medidas radiográficas clínicas padrões (Capítulo III, Estudo 3.3).

(iv) Investigar a influência do tipo de aterrisagem na corrida - em antepé ou retropé - no comportamento cinemático do tornozelo e pé durante a fase de apoio (Capítulo IV, Estudo 4.1).

(v) Investigar a relação do padrão cinemático do tornozelo e pé durante a aterrissagem na corrida de antepé e as forças verticais e taxas de carga (Capítulo IV, Estudo 4.2).

(vi) Investigar a eficácia de uma intervenção terapêutica para os pés na cinemática do pé e nas forças e taxas de carga de corredores recreacionais (Capítulo II, Estudo 2.1 e Capítulo V, Estudo 5.1).

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CAPÍTULO II - PROTOCOLO

DO ENSAIO CLÍNICO

Neste capítulo é apresentado o artigo do protocolo desenvolvido para avaliar a eficácia de uma intervenção terapêutica para os pés, na cinemática do pé e nas forças e taxas de carga de corredores recreacionais de longa distância.

2.1 Protocol for evaluating the effects of a therapeutic foot exercise program on injury incidence, foot functionality and biomechanics in long-distance runners: a randomized controlled trial

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STUDY PROTOCOL Open Access

Protocol for evaluating the effects of a therapeutic foot exercise program on injury incidence, foot functionality and biomechanics in long-distance runners: a randomized controlled trial

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Abstract

Background: Overall performance, particularly in a very popular sports activity such as running, is typically influenced by the status of the musculoskeletal system and the level of training and conditioning of the biological structures. Any change in the musculoskeletal system's biomechanics, especially in the feet and ankles, will strongly influence the biomechanics of runners, possibly predisposing them to

injuries. A thorough understanding of the effects of a therapeutic approach focused on feet biomechanics, on strength and functionality of lower limb muscles will contribute to the adoption of more effective therapeutic and preventive strategies for runners. **Methods/Design**: A randomized, prospective controlled and parallel trial with blind assessment is designed to study the effects of a "ground-up" therapeutic approach focused on the foot-ankle complex as it relates to the incidence of running-related injuries in the lower limbs. One hundred and eleven (111) healthy long-distance runners will be randomly assigned to either a control (CG) or intervention (IG) group. IG runners will participate in a therapeutic exercise protocol for the foot-ankle for 8 weeks, with 1 directly supervised session and 3 remotely supervised sessions per week. After the 8-week period, IG runners will keep exercising for the remaining 10 months of the study, supervised only by web-enabled software 3 times a week. At baseline, 2 months, 4 months and 12 months, all runners will be assessed for running-related injuries (primary outcome), time for the occurrence of the first injury, foot health and functionality, muscle trophism, intrinsic foot muscle strength, dynamic foot arch strain and lower-limb biomechanics during walking and running (secondary outcomes). **Discussion**: This is the first randomized clinical trial protocol to assess the effect of an exercise protocol that was designed specifically for the foot-and-ankle complex on running-related injuries to the lower limbs of long-distance runners. We intend to show that the proposed protocol is an innovative and effective approach to decreasing the incidence of injuries. We also expect a lengthening in the time of occurrence of the first injury, an improvement in foot function, an increase in foot muscle mass and strength and beneficial biomechanical changes while running

and walking after a year of exercising. **Trial registration:** Clinicaltrials.gov Identifier NCT02306148 (November 28, 2014) under the name "*Effects of Foot Strengthening on the Prevalence of Injuries in Long Distance Runners*". Committee of Ethics in Research of the School of Medicine of the University of Sao Paulo (18/03/2015, Protocol # 031/15).

Keywords: running, sports injuries, exercise therapy, foot, biomechanics

Background

Human performance, particularly in one of the most popular sports activities such as running, is typically influenced by the state of the musculoskeletal system, either by the level of training and conditioning of the biological structures, or by the aging process. Although popular worldwide due to its low cost, versatility, convenience [1], and health benefits to people of all ages [2], running is associated with a high prevalence of lower extremity injuries (between 19.4% and 79.3%) [3]. The occurrence of injuries limits the intended benefits by inducing changes in practice habits [4] or temporary or even permanent cessation of running. In addition, injuries lead to increased costs due to medical treatment and/or work absence [5].

The understanding of risk factors associated with these injuries, particularly the intrinsic factors, can provide important benefits for runners. Among these intrinsic factors, those that are noteworthy include biomechanical factors and muscle functionality of the lower extremities, particularly the feet. A systematic review by van der Worp et al. [5] included 11 high-quality longitudinal studies and concluded that alterations in the biomechanical force distribution patterns, amount of training, history of previous injuries, increased index of the navicular drop, and the misalignment of the ankle, knee, and hip are among the main intrinsic risk factors for running-related injuries. In addition, extrinsic factors such as the training surface and the type of footwear are also relevant risk factors [5]. It is noteworthy that out of these seven diverse risk factors, two are related to the foot-ankle complex, demonstrating the importance of maintaining the health and functionality of its musculoskeletal structures to prevent injuries. It is also believed that any biomechanical alteration in the musculoskeletal system, in particular the foot-ankle complex, broadly influences a runner's functionality, predisposing him/her to a lesser or greater extent to injuries, in addition to the possibility of compromising his/her quality of life [2,6].

The foot has a complex structure that can perform a broad variety of functions in different postural and dynamic tasks [7,8]. This versatility can only be achieved through its unique arch-shaped architecture and its powerful intrinsic and extrinsic muscular activity, which is responsible for the maintenance and control of foot arches, postural corrections during disturbances, and torque generation during body displacement [9,10]. Even with this unique and specialized structure, a high prevalence of injuries associated with running practices occurs in this complex. Among the most common hypotheses used to explain this high prevalence are factors such as the excessive ankle/foot pronation in the stance phase of running [11], the lowering of the medial longitudinal arch due to navicular drop [12,13], the alteration of rigidity of the plantar arches [14], and the increase in impact and acceleration of the tibia during running [15].

Evidence suggests the importance of the intrinsic foot musculature, showing that fatigue can cause a significant increase in pronation, which is evaluated by the navicular drop [12]. In addition, weakness may be a risk factor for falls in the elderly population [16]. Therefore, it is understandable that the specific training of foot [13,17] and ankle muscles [18–20] is an important tool that improves functions and functionalities of the lower extremities, as has been shown in recent studies [13,19–21].

In one of those studies, the unsupervised practice of a single exercise for the feet (short-foot exercise) four times a week promoted a decrease in the navicular drop, an increase in the medial longitudinal arch index, and an increase in the functionality quality of the intrinsic foot muscles in asymptomatic individuals [13]. These results were maintained one month after the training had been completed. Although the results of Mulligan and Cook [13] are promising, they only measured the foot function in static conditions and the unsupervised practice of an isolated exercise for four weeks may not have been sufficient to cause a transfer of the static gains for a more dynamic task where the foot would be more robustly utilized, according to the star excursion balance test. In contrast, one study compared two groups: one group performed a four-week period of short-foot exercises, including 100 repetitions for five seconds each, and the second group performed a four-week period of towel-curl exercises with the same amount of exercise [20]. This controlled study showed that both groups exhibited decreased displacement of the centre of pressure during the modified star excursion test. Therefore, a load increase in the same exercises used by Mulligan and Cook [13] resulted in positive effects for postural control.

The same short-foot exercise was practiced by individuals with flat feet in a randomized controlled trial to investigate its effect on the use of foot orthoses [17]. The protocol consisted of three to five sets of exercises with five repetitions each, twice a day, for eight weeks. In both study groups, the isometric force and the transversal section area of the abductor hallucis muscle were increased after the interventions, with a significant increase in the group that used orthoses during exercises. These results demonstrated that even in structurally unfavourable conditions, exercise for the foot muscles leads to important strength gains. It is noteworthy that even with a well-planned intervention, the lack of a control group and the evaluation of the muscle strength alone limit the study conclusions. In addition, the study did not take into account the potential clinical and functional changes of the plantar arches, as performed by Goldmann et al. [19]. This group of researchers investigated the effects of the hallucis flexors strengthening in the kinetic and kinematic of foot and ankle during walking, running, and vertical jumping among university athletes. Training of the experimental group consisted of isometric contractions of the hallucis flexors at 90% of the maximum voluntary contraction using a dynamometer four times a week for seven weeks. The authors observed a significant increase in the performance of vertical jumping and extensor and flexor momentum of the metatarsal-phalangeal joint and a gain of 60% to 70% in the strength of the hallucis flexors. This study shows that the flexor muscles of the foot respond in a quick and intense manner to training, even for simple training, the strengthening of the muscles in question results in global kinematic and kinetic alterations. It would still be interesting to determine how long these gains would last after the completion of the intervention and whether more elaborate training, involving more muscles and different postures and loads, would alter the study outcome, especially regarding to the foot biomechanics during locomotor tasks.

The understanding of the effects of a therapeutic approach focused on the foot biomechanics of walking and running, on the strength and functionality of lower extremity muscles will contribute to the adoption of more effective therapeutic and preventive strategies for runners. However, no evidence exists that supports the efficacy of the therapeutic exercises already used and recommended for the health of the feet [7,17,19,20,22] to prevent recurrent injuries in long-distance runners. However, one research protocol aims to assess the effects of ankle and hip muscle strengthening and functional balance training on running mechanics, postural control, and injury incidence in novice runners with less than one year of running experience but without focusing on the intervention of intrinsic and extrinsic muscles of the feet [23].

Therefore, a controlled and randomized clinical trial would determine whether these interventions are efficacious by using the incidence of running-related injuries as the primary outcome and following both intervention and control subjects during a period equal to or greater than one year (the period during which the incidence and prevalence of these injuries are reported) [4,24–28].

It is important to highlight that rehabilitation programs rarely include the intrinsic muscles of the feet in their therapeutic protocols. The present proposal uses a new paradigm in which the focus of training and preventive interventions in runners is a “ground-up” approach rather than the traditional "top-down" approach, which focuses on the hip strengthening. This new approach, advocated by Baltich et al. [23], will seek to improve the function of the ankle-foot complex, which is directly associated with the absorption and transmission of body forces to the ground and vice-versa during running.

Hypotheses

Our hypotheses are that the therapeutic exercise protocol for the foot-ankle as practiced by long-distance recreational runners for one year will:

H 1. Reduce the incidence of running-related injury in the lower limbs,

H 2. Lengthening the time for the occurrence of the first running-related injury in the lower limbs,

H 3. Increase intrinsic foot muscle strength,

H 4. Increase foot muscle cross-sectional area and volume,

H 5. Improve foot health and functionality status,

H 6. Reduce dynamic strain on the foot’s longitudinal arch during running and walking, and

H 7. Produce beneficial biomechanical changes during running that denote an improvement in the mechanical efficiency of absorbing loads and propelling the body while walking and running. Such changes would include an increase in the ankle range of motion in the sagittal plane and increases in 1) ankle extensor moment and power and 2) knee extensor moment and power during the second half of the stance phase.

Our aim is therefore to investigate the effects of a "ground-up" therapeutic approach focused on the foot-ankle for one year as they relate to 1) the incidence of running-related injuries in the lower limbs of long-distance runners, 2) time of occurrence of the first injury, 3) foot health and functionality, 4) strength of the intrinsic foot muscles; 5) foot muscle trophism, 6) dynamic foot arch strain and 7) lower-limb biomechanics during walking and running.

Methods/Design

Overview of the Research Design

A randomized, prospective controlled and parallel trial with blind assessment is designed to study the effects of a "ground-up" therapeutic approach focused on the foot-ankle concerning the incidence of running-related injuries to the lower limbs of long-distance runners. This trial has an allocation ratio of 1:1. Its framework is exploratory to gather preliminary information on the intervention of conducting a full-scale trial. The trial follows all recommendations established by SPIRIT [29].

Long-distance recreational runners are recruited from the vicinity of the city of São Paulo and referred to a physical therapist, who performs the group allocation. The participants are then referred to another physical therapist, who performs the initial blind assessment. All runners allocated to the intervention group (IG) participate in a protocol of therapeutic exercises for the foot-ankle complex for 8 weeks, with one session per week supervised by a physical therapist and three sessions per week remotely supervised by web-enabled software [30]. They receive access to the web software on the first day and use it for 8 weeks. After the 8-week period, the IG runners will continue exercising for 10 more months, supervised only by the web software 3 times a week. The runners allocated to the control group (CG) do not receive any intervention training but receive a placebo stretching exercise program.

All runners will be assessed at baseline and 2 months (end of intervention). They are then assessed twice more for follow-up purposes, at 4 and 12 months after the baseline. Assessments will concern the incidence of running-related injuries (primary outcome), and all other secondary outcomes.

The design and flowchart of the protocol are presented in Figure 1. The assessments are performed at the Laboratory of Biomechanics of Human Movement and Posture (LaBiMPH) at the Physical Therapy, Speech and Occupational Therapy department of the School of Medicine of the University of São Paulo, São Paulo, Brazil.

Participants and Recruitment

This study is currently recruiting patients (study start date: April 2015)

The eligibility criteria for the volunteer runners are:

- aged between 18 and 55 years old
- at least one year of running experience

- a weekly training distance greater than 20 km and less than 100 km as their main physical activity
- within two months prior to baseline assessment, lack of any lower limb musculoskeletal injury or pain that might lead to stopping running practice
- no prior experience within the last year of isolated foot and ankle strength training
- not receiving any physical therapy intervention
- no history of using minimalist shoes for running practice
- no prior experience of barefoot running

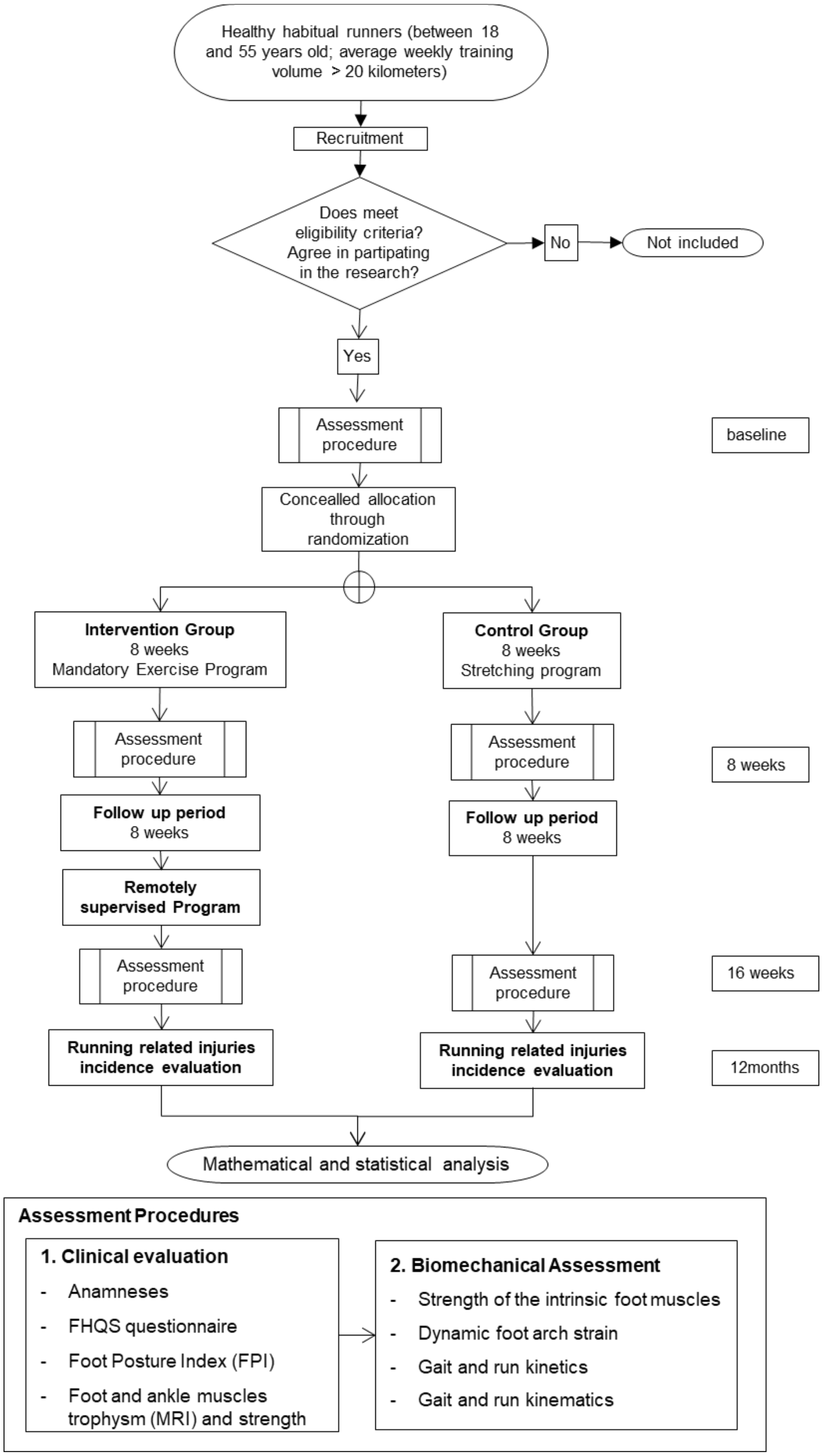

Figure 1 - Flow chart of the study's design.

Runners are not selected if they have other neurological or orthopedic impairments (such as congenital foot malformations, stroke, cerebral palsy, poliomyelitis, rheumatoid arthritis, prosthesis or moderate or severe osteoarthritis), major vascular complications (venous or arterial ulcers), diabetes mellitus, sequelae from poorly healed fractures or prior lower-limb surgeries.

These runners may use the running technique of fore-, mid- or rear-foot ground contact, which will be classified by the strike index, according to Cavanagh and Lafortune [31].

One hundred and eleven (111) runners will be recruited by radio advertisements, print media and running association groups at their site of practice around the city of São Paulo. The potential subjects will be interviewed by telephone and, when selected, assessed in the laboratory to confirm all the eligibility criteria. This first laboratory assessment represents the baseline condition (blind assessment).

The runners allocated to the IG will be treated during their locally supervised session at the Physical Therapy Department in an ambulatory setting that assists all the physical therapy treatments of the Department, providing a reliable therapeutic environment for the intervention.

Randomization, Allocation and Blinding

The randomization schedule was prepared using Clinstat software [32] by an independent researcher (Researcher 1) who was not aware of the numeric code for the CG and IG groups. A numeric block randomization sequence will be kept in opaque envelopes.

After the runners' agreement to participate and assignment in the research, the allocation into the groups will be made by another independent researcher (Researcher 2), who also will be unaware of the codes. Only the physiotherapist (Researcher 3) responsible for the locally supervised training knows who is receiving the intervention. Researcher 3 will also be responsible for the remote monitoring of the training by web software [30] and telephone. One physiotherapist (Researcher 4), who will also be blind to the treatment allocation, will be responsible for all clinical, functional, and biomechanical assessments. Both physiotherapists (researchers 3 and 4) will be blind to the block size used in the randomization procedure.

To guarantee the blindness of researcher 4, before each evaluation, runners will be instructed not to reveal whether they are in the CG or IG; their questions should be asked only to the physiotherapist in charge of web software [30] and local training (Researcher 3).

The trial statistician will also be blind to treatment allocation until the main treatment analysis has been completed.

Treatment Arms

The CG runners will receive a 5-minute placebo routine of warm-up and muscle-stretching exercises to be performed immediately before every running practice during their 8-week study (Additional File 1, table 3).

The IG runners will receive a therapeutic foot-ankle exercise protocol for strengthening and improving functionality under the supervision of a physiotherapist

(Researcher 4) once a week for 8 weeks, and a series of foot-ankle exercises to be performed under remote supervision through web software [30] 3 times a week for the full 1-year length of the study (1 year). Both locally (Additional File 1, table 1) and remotely supervised therapeutic routines (Additional File 1, table 2) will take from 20 to 30 minutes. In particular, the remotely supervised practice will be preferentially performed at home; the web software includes written descriptions, photos and videos of each exercise.

Each week, IG runners will be requested to evaluate the subjective effort of each exercise's performance using a score of 0 to 10 either with the web software [30] or to the physiotherapist during locally supervised practice. If the effort score ranges from 0 to 5 and the runner's performance of each exercise is found adequate during the supervised session by the physiotherapist, the exercises will increase in difficulty according to the progression chart in tables 1 and 2. If the effort score ranges from 6 to 7, the exercise will not increase in difficulty and no progression would be done on that exercise. Thus, the runner remains in the same exercise progression until he/she scores 0 to 5 in each particular exercise. Finally, if an IG runner reports a score from 8 to 10, the exercise will decrease in difficulty, if possible, until the subject is able to perform it without pain or discomfort.

Assessments

A physiotherapist (Researcher 3) who is blind to group allocation will perform all assessments. Each assessment will consist of taking a clinical history of personal details, anthropometry, running practice details (years of practice, weekly frequency and volume, usual shoe and training surface, number of races and whether the runner trains with a running coach), previous orthopedic surgery, other physical activity practiced regularly (previous to running practice or simultaneously with running) and an injury history concerning the most important risk factors previously published [3,33,34].

A foot-health status questionnaire [35] will be used to characterize foot health and functionality. We will use a Brazilian-Portuguese version (FHSQ-BR) translated and validated by Ferreira et al. [36]. This instrument is divided into three sections. Section I evaluates foot health in four domains: foot pain, foot function, footwear, and general foot health. Section II evaluates general health in four domains: general health, physical activity, social capacity and vigour. Sections I and II are composed of questions with answer options presented in affirmative sentences and corresponding numbers. Section III collects general demographic data of the individuals [36]. We will not use the scores from Section III. Each domain scores from 0 to 100 points, where 100 is the best condition and 0 the worst.

We will access variations in foot posture of the runners using the Foot Posture Index (FPI) [37]. The FPI is a six component measures that allows multiple segment evaluation of foot posture on a static measurement and requires that subjects stand in their relaxed stance position looking straight ahead while the assessment is in process. The assessment consists of the (1) palpation of the talar head, (2) observation of supra and infra malleolar curvature, (3) observation of the calcaneal frontal plane position, (4) observation of the bulging in the region of the talo-navicular joint, (5) observation of the height and congruence of the medial longitudinal arch and (6) presence of abduction or adduction of the forefoot. Scores reaching from -12 to +12 and normative values are presented on the literature.

Subjects will then be assessed for intrinsic foot muscles strength, lower-limb running kinematics and kinetics, and dynamic foot-arch strain. The feet of 30% of the participants in each group (41 participants) will be imaged by magnetic resonance imaging (MRI) to assess trophism and strength of the foot intrinsic muscles; this will be scheduled for the same week of each subject's baseline measurements.

After baseline assessment, all subjects will be scheduled for 2 follow-ups assessments, one at 8 weeks and the other at 16 weeks. They will maintain contact with the Researcher 3 through the follow-up period by the web software [30], e-mail and telephone.

Running-related Injuries

Running-related injuries will be assessed initially at the baseline and will be assessed continually throughout the study by the web software [30]. The definition of running-related injury was set according to the study of Macera et al. [4]. They stated that any musculoskeletal pain or injury that was caused by running practice and that induces changes in the form, duration intensity or frequency of training for at least one week will be considered a running-related injury. Only lower-limb injuries will be accounted during the 12-month period after the baseline assessment; both the incidence and time of occurrence of the first injury will be analyzed.

If any subject presents a new injury during his or her participation in the study, the injury will be accounted for and the intervention or placebo intervention will be discontinued, even though all subjects will keep being followed for the completion of the study.

Isometric Intrinsic Foot Muscles Strength

Strength of the foot's intrinsic muscles will be assessed in trials using a pressure platform (EMED: Novel, Germany) on which the subjects will place their dominant foot while standing with knees extended. They will push down as hard as possible using only their hallux and toes, particularly the metatarsophalangeal joints and not the hallux interphalangeal joint. A physiotherapist will determine whether the subject lifted the heel and inspect fluctuations in the line of gravity and trunk posture during each trial. If any changes are observed in the line of gravity or positioning of the heel or trunk, the trial will be excluded. Three trials will be completed on each foot (left and right) according to Mickle et al. (2006)[38]. Maximum force will be normalized by body weight and analyzed for hallux and toes areas separately.

Foot Muscle Trophism and Strength

One indirect method of measuring foot strength is through MRI, which, combined with other techniques, offers good reliability and a way to follow changes in muscular volume [39]. In addition, MRI can facilitate understanding the etiology of running-related injuries and rehabilitation of the foot-ankle complex [40].

The MRI of the foot will be performed with a 1.5T system. Foot images will be acquired by the same technician using a coil of four channels positioned in the magnetic centre. Participants will be placed in supine position with the ankle at 45°

of plantar flexion inside the coil. Images will be acquired in the frontal, sagittal and transverse planes to confirm the position of the feet, and the subject will be repositioned if necessary. T1-weighted images of the entire foot length will be acquired perpendicular to the plantar aspect of the foot using a spin-echo sequence (repetition time=500 ms, echo time=16 ms, averages=3, slice thickness= 4 mm, gap between slices=0 mm, field of view=120×120 mm, flip angle=90°, matrix=512×512) [41]. The set of images will cover the distance between the most proximal and most distal images in which every intrinsic foot muscle is visible.

To assess changes in the cross-sectional area (CSA) and volume of the intrinsic foot muscles, 30% of the subjects from each group will have MRI of the foot at three times: baseline, 8 weeks and 16 weeks.

The CSA will be measured by ImageJ planimeter software [42]. Following, Miller et al. [14] for each muscle at each slice and muscle volume will be calculated by multiplying the CSA of all slices for a muscle by their linear distance (4mm) and adding these volumes.

Walking and Running Biomechanics

To ensure maximum reliability, all biomechanical testing sessions will be completed by the same researcher.

Gait and running kinematics will be acquired using three-dimensional displacements of passive reflective markers (10 mm in diameter) tracked by nine infrared cameras at 100 Hz (OptiTrack FLEX: V100, Natural Point, Corvallis, OR, USA) [43,44]. Some 14 markers will be placed on the right subject's foot according to

Leardini's protocol [45]. Extra markers will be placed at the medial knee joint line, lateral knee joint line and bilaterally at the iliac spine antero-superior, superior aspect of the greater trochanter, and sacrum. These markers will be used to determine relative joint centres of rotation for the longitudinal axis of the foot, ankle and knee. The extra markers from the medial aspect of the knee joint line will be removed during the dynamic trial. In addition, three non-collinear reflective markers will be fixed at two technique clusters. One of the clusters will be placed in the lateral thigh and the other over the shank.

The laboratory coordinate system will be established at one corner of the force plate and all initial calculations will be based on this coordinate system. Each lower-limb segment (shank and thigh) will be modelled based on surface markers as a rigid body with a local coordinate system that coincides with the anatomical axes. Translations and rotations of each segment will be reported relative to the neutral positions defined during the initial static standing trial. All joints will be considered spherical (i.e., with three rotational degrees of freedom). The foot will be modeled according to Leardini et al. [45]. That is, the calcaneus, mid-foot and metatarsus are considered rigid bodies and the longitudinal axis of the first, second and fifth metatarsal bones and proximal phalanx of the hallux will be tracked independently.

Ground reaction forces will be acquired by a force plate (AMTI OR-6-1000, Watertown, MA, USA) with a sampling frequency of 1kHz embedded in the centre of the walkway. Force and kinematic data acquisition will be synchronized and sampled by an A/D card (AMTI, DT 3002, 12 bits).

The subjects will go through a habituation period before the data acquisition to establish confidence and comfort in the laboratory environment, and to ensure appropriate movement velocity. To assess lower extremity running mechanics, subjects will perform 10 valid over-ground walking trials and 10 valid over-ground running trials at a constant velocity (9.5 km/h to 10.5 km/s); these will be monitored by two photoelectrical sensors (Speed Test Fit Model, Nova Odessa, Brazil).

The automatic digitizing process, 3D reconstruction of the markers' positions and filtering of kinematic data will be performed using AMASS software (C-motion, Kingston, ON, Canada). Kinematic data will be processed using a zero-lag second-order low-pass filter with cutoff frequencies of 6Hz for walking and 12 Hz for running**.** Ground reaction force data will be processed using a zero-lag low-pass Butterworth fourth-order filter with cutoff frequencies of 50Hz for walking and 200 Hz for running.

A bottom-up inverse dynamics method will be used to calculate the net moments in the sagittal and frontal planes of the ankle and knee joints using Visual3D software (C-motion, Kingston, ON, Canada). The human body will be modeled by three linked segments (foot, shank and thigh) and the inertial properties will be based on Dempster's standard regression equations. The moment of inertia and location of center of mass will be computed assuming the thigh and shank segments as cylinders.

Calculation of all variables will be performed using a custom-written MATLAB function (MathWorks, Natick, MA, USA). Data of only one lower limb (randomly chosen) per subject will be analyzed and compared.

The following ankle kinematic variables will be analysed: maximum dorsiflexion at foot contact, maximum plantarflexion, maximum dorsiflexion at the toe-off and dorsiflexion range of motion (ROM) in the sagittal plane during the stance phase. The knee kinematic variables are: maximum flexion at foot contact, maximum extension, maximum flexion in the stance phase, ROM on sagittal plane, maximum abduction and adduction in the stance phase. The foot kinematic variables are: elevation/drop of the longitudinal arch angle and of the first, second and fifth metatarsal bones; rearfoot to forefoot rotation; transverse plane angle between first and second metatarsal bones and between second and fifth metatarsal bones; and maximum inversion and eversion of the calcaneus (frontal plane).

The ankle and knee kinetic variables to be analysed are net ankle and knee moments normalized by body weight times height and power normalized by body weight in the sagittal plane. The ground reaction force variables will be normalized by body weight and are as followings: first peak force (body weight – BW), second peak (BW), loading rate 80 [N/ms], defined as the force rate between 20 and 80% of the contact of the foot with the ground during the first peak; loading rate 100 [N/ms], as determined by the force rate between 0 and 100% of the first peak and push-off rate [N/ms], as defined as the rate of the second peak force, between the minimal values until the second peak.

Dynamic longitudinal foot arch strain

The dynamic longitudinal foot arch strain will be measured according to Liebermann et al. [46]. The measurement involves navicular height (NH), which is the minimum distance from the navicular tuberosity relative to the line formed by the first metatarsal head and the medial process of the calcaneus. These three landmarks form a plane and NH is independent of rear-foot inversion or eversion. Arch strain can also be quantified by fitting a parabola to markers 1–3 (with the navicular head as the vertex) and then measuring the average curvature at 100 points evenly spaced along the curve.

Outcome Measurements

The primary outcome measurement will be the incidence of running-related injuries in the lower limbs accounted at the end of 12 months of study.

The secondary outcomes will be: 1) the time of the occurrence of the first injury along the study period (time to event); 2) foot health and functionality (change from baseline); 3) foot, ankle and knee kinematics, ankle and knee joint moments, and knee and ankle power during walking and running (change from baseline); 4) strength of the intrinsic foot muscles (change from baseline); 5) foot muscle trophism (change from baseline); and 6) dynamic foot arch strain (change from baseline).

Interventions

Runners allocated to the IG will receive a foot-ankle therapeutic exercise protocol for strengthening and improving functionality. Part of the exercise protocol (12 exercises) is to be performed once a week under the supervision of a physiotherapist for 8 weeks (Additional File 1, Table 1). And a series of eight foot-ankle exercises is also to be performed 3 times a week remotely supervised by web software [30] (Additional File 1, Table 2) for the full 1-year completion time of the study. Each session, whether supervised locally or remotely, lasts 20 to 30 minutes. The therapeutic exercise protocol is described in detail in tables 1 and 2.

Gradual and progressive difficulty will be offered to the runner, respecting any limitation due to pain, fatigue and/or decrease in performance during execution. The runners in the IG will access the web software [30] daily, entering their data regarding performance of the foot exercise training and ranking their level of difficulty in each exercise from 0 to 10.

During the locally supervised sessions, the physiotherapist will focus on proper alignment of the foot-ankle segments, especially if the runner has difficulty in maintaining it, in a way that allows no movement compensations.

Runners allocated to the CG will receive a 5-minute placebo warm-up and muscle stretching exercise routine (Additional File 1, Table 3) that they are to perform for 8 weeks immediately before each running practice. This placebo training can also be assessed and followed through the web software [30]. The stretching exercises are described in detail in Table 3. We hypothesized that a warm-up combined with muscle stretching exercises would not have any effect on foot muscular strength and functionality, lower extremity biomechanics or injury prevention.

Both groups will access the web software [30] daily, entering their running practice data (daily training duration and volume) and information concerning the occurrence of any injury event.

The discontinuation criteria for the exercises during any session includes cramps, moderate to intense pain, fatigue or any other condition that exposes the runner to any discomfort.

The discontinuation criteria for the training includes an occurrence of a running-related injury in the lower limbs.

If any subject fails to access the web software [30] for three consecutive weeks without explanation or fails to attend the locally supervised training three consecutive times, that subject will be terminated from the study.

To improve adherence, several actions will be performed by the researchers in the web software [30]. Data regarding the subjects running practice, such as training volume, time of practice and occurrence of injuries, will be reported to the web software, which will summarize it and make it viewable in the users' area. In addition, for the duration of the study, runners' responses in the web software concerning their foot-ankle exercise practice and running training will be stored and be accessible to the researchers and subjects at any time. If any subject fails to log in to the web software for more than 5 consecutive days, an e-mail will automatically be sent, asking the subject to log in to his or her account and report data on the training (or lack of it) for the past week. The physiotherapist responsible for the therapeutic protocol will make phone contact with subjects who fail to attend to any of the weekly locally supervised sessions. They will also make phone contact with subjects who do not respond to e-mail reminders from the web software. Subjects will also be contacted by personal phone calls if data they reported on the web software is found to be inconsistent [47].

After the period of intervention and after 12 weeks of follow up all runners will be questioned about their satisfaction to the training protocol with one question (Did you enjoy doing the exercises?) with three answer possibilities (No; A Little; A lot). To avoid evaluation bias, runners will answer this question secretly through an online-unidentified form sent to their e-mail. Runners will be informed about the anonymity and this form will only be accessed after completion of the study.

For the duration of the trial, subjects will be advised not to engage in any new physical activity or preventive training protocols for the foot and ankle. If any subject cannot avoid such behavior, he or she must report this situation during web software [30] access. Concomitant care, such as physical therapy, acupuncture or other conventional medical care, will not be permitted except for runners who are injured during the study. At the end of 12 months, CG participants that are interested will receive access to the software for the foot exercise protocol.

Sample size and statistical analysis

The sample size calculation was made using an effect size of 0.28 (proportion), considering the categorical primary outcome variable, which is the incidence of running-related lower-limb injuries [34]. A sample size of 101 runners is needed to provide 80% power to detect a moderate effect difference between the highest and lowest group injury incidence medians, assuming an alpha of 0.05 and a $\chi 2$ statistical design – contingency tables (df=1) [48]. Assuming a 10% dropout rate during the study, a sample size of 111 runners is needed.

The statistical analysis will be based on intention-to-treat analysis, and mixed general linear models of analysis of variance for repeated measure will be used to detect treatment-time interactions (α = 5%). The outcome measures will be compared at baseline, 2 months, 4 months and 12 months. Effect sizes (Cohen´s d coefficient) will also be provided between baseline and 2 months and between 2 months and follow-up (4 and 12 months), if the intervention shows any treatment effect. The missing data will be treated by imputation methods depending on the type of the missing data we will face: missing completely at random, missing at random, or missing not at random [49].

Ethics and Data security

This trial was approved by the Ethics Committee of the School of Medicine of the University of São Paulo (Protocol number nº031/15). Additionally, this trial is registered in ClinicalTrials.gov (a service of U.S. National Institutes of Health) Identifier NCT02306148 (November 28, 2014) under the name "*Effects of Foot Strengthening on the Prevalence of Injuries in Long Distance Runners".* All runners will be asked for written informed consent according to the standard forms and the researcher 4 will obtain them. All personal data from potential or enrolled runners will be maintained confidential before, during and after the trial by encoding participant's name. All data access and storage are in keeping with National Health and Medical Research Council guidelines, as approved. All files will be available from the database published at figshare.com. All-important protocol amendments will be reported to investigators, review boards and trial registration by the Researcher 3.

Discussion

This clinical trial will provide important data on foot-training effectiveness, its influence on the incidence of injuries and its efficacy on strengthening the muscles of the foot-ankle complex. It will also facilitate the identification of risk factors and biomechanical mechanisms involved in injury processes and prevention. We also intend to contribute new evidence that could be used as a guide for further studies on biomechanical changes in dynamic tasks resulting from the strengthening of the foot-ankle complex.

The few existing clinical trials that have proposed exercise protocols to reduce the incidence of runners' injuries have not included the incidence of injury as a primary outcome. They also have had short follow-up periods and usually failed to follow the subjects' adherence to the program and the correctness of exercise performance throughout the study [13, 17, 19, 20]. In contrast, this trial has the incidence of running-related injuries as a primary outcome, will have a long period of follow-up (12 months), proposes an intervention training protocol with several exercises that are easy to perform with short durations for each session (20-30 minutes) and does not require subjects to be continuously supervised by a health professional. In addition, it utilizes open-access web software [30] that will support adherence control.

We understand that the number of MRIs that we are performing (on 30% of the subjects) is limited and might prevent a broad conclusion about changes in intrinsic foot muscle cross-sectional area (CSA) and volume.

Running-related injuries in this population cause interruptions and abandonment of physical activity. They also could lead to the development of chronic injury that would prevent the practice of other sports and hence frustrate the individual's pursuit of a healthy lifestyle. Runners are constantly looking for ways to remain free from injury and the information they receive from coaches or media is often conflicting and varied [50]. Our protocol has the potential to change the course of this vicious cycle experienced by long-distance runners.

If our hypothesis that such an exercise protocol reduces the incidence of running-related injuries to long-distance runners is confirmed, it could be easily incorporated into their warm-up routine prior to running practice.

Declarations

List of abbreviations

CG - Control group

IG - Intervention group

FPI - Foot Posture Index

FHSQ-BR - Foot-health status questionnaire - Brazilian-Portuguese version

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Ethics approval and consent to participate

This trial was approved by the Ethics Committee of the School of Medicine of the University of São Paulo (Protocol number nº031/15). Additionally, this trial is registered in ClinicalTrials.gov (a service of U.S. National Institutes of Health) Identifier NCT02306148 (November 28, 2014) under the name "Effects of Foot Strengthening on the Prevalence of Injuries in Long Distance Runners". All runners will be asked for written informed consent according to the standard forms and the researcher 4 will obtain them.

Authors' contributions

All authors have made substantial contributions to all three of sections (1), (2) and (3): (1) The conception and design of the study, or acquisition of data, or analysis and interpretation 558 of data (2) drafting the article or revising it critically for important intellectual content (3) final approval of the version to be submitted. And in the protocol the following roles will be played by the authors: UTT is responsible for the study design, intervention, interpretation of the data, writing the report and submission of the manuscript. ABM is responsible for the study design, data collection, management, analysis, and interpretation, writing the report and submission of the manuscript. ICNS is responsible for the study design, interpretation of the data, writing the report 564 and submission of the manuscript.

Competing interests

The authors affirm that this study has not received any funding/assistance from a commercial organization which may lead to a conflict of interests.

Availability of data and materials

All personal data from potential or enrolled runners will be maintained confidential before, during and after the trial by encoding participant's name. All data access and storage are in keeping with National Health and Medical Research Council guidelines, as approved. All files will be available from the database published at figshare.com. All-important protocol amendments will be reported to investigators, review boards and trial registration by the Researcher 3.

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CAPÍTULO III - MODELO MULTISEGMENTAR DO PÉ PARA AVALIAÇÃO BIOMECÂNICA

Esse capítulo inclui três artigos originais que testaram e discutiram a confiabilidade e a usabilidade de um modelo multisegmentar do pé para avaliar a cinemática de suas articulações na corrida. Adicionalmente, este modelo foi aprimorado e teve sua confiabilidade e acurácia testadas para a avaliação do arco longitudinal plantar e sua deformação, tomando como base a anatomia do arco.

3.1 Repeatability of skin-markers based kinematic measures from a multi-segment foot model in walking and running

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Repeatability of skin-markers based kinematic measures from a multi-segment foot model in walking and running

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Abstract

Skin-markers based multi-segment models are growing in popularity to assess foot joint kinematics in different motor tasks. However, scarce is the current knowledge of the effect of high-energy motor tasks, such as running, on the repeatability of these measurements. This study aimed at assessing and comparing the inter-trial, inter-session, and inter-examiner repeatability of skin-markers based foot kinematic measures in walking and running in healthy adults. The repeatability of 24 kinematic measures from an established multi-segment foot model were assessed in two volunteers during multiple barefoot walking and running trials by four examiners in three sessions. Statistical Parametric Mapping (1D-SPM) analysis was performed to assess the degree of shape-similarity between patterns of kinematic measurements. The average inter-trial variability across measurements (deg) was 1.0±0.3 and 0.8±0.3, the inter-session was 3.9±1.4 and 4.4±1.5, and the inter-examiner was 5.4±2.3 and 5.7±2.2, respectively in walking and running. Inter-session variability was generally similar between the two motor tasks, but significantly larger in running for two kinematic measures ($p<0.01$). Inter-examiner variability was generally larger than inter-trial and inter-session variability. While no significant differences in frame-by-frame offset variability was detected in foot kinematics between walking and running, 1D-SPM revealed that the shape of kinematic measurements was significantly affected by the motor task, with running being less repeatable than walking. Although confirmation on a larger population and with different kinematic protocols should be sought, attention should be paid in the interpretation of skin-markers based kinematics in running across sessions or involving multiple examiners.

Keywords: foot kinematics; skin-markers; walking; running; repeatability; errors.

Introduction

The foot is a complex biomechanical structure with multiple degrees of freedom. In order to measure foot joint motion non-invasively, a large number of skin-markers based kinematic protocols have been implemented and reported in the literature (Leardini et al., 2019). Diagnosis of musculoskeletal pathologies (Khazzam et al., 2007; Rao et al., 2007; Chang et al., 2014; Deschamps et al., 2016) quantitative assessment of footwear and foot orthotics (Barton et al., 2011; Oosterwaal et al., 2011; Leardini et al., 2014; Bishop et al., 2016) and evaluation of sport tasks' performances (Arndt et al., 2007; Kelly et al., 2014; Takabayashi et al., 2018) are only few examples of the importance of these kinematic protocols across several research fields. Their applications have been further boosted by the sport biomechanics community, due to the increasingly large popularity of recreational running across age-groups and populations worldwide. Among these protocols, the capability to track the midfoot segment have helped increase the applications of the Rizzoli Foot Model (RFM) (Leardini et al., 2007; Deschamps et al., 2012b; Portinaro et al., 2014) also outside the clinical context (Leardini et al., 2019). Despite its extensive use in running biomechanics (Powell et al., 2013; Shih et al., 2014; Sinclair et al., 2014, 2015; Sterzing et al., 2015; Trudeau et al., 2017; Kelly et al., 2018; Langley et al., 2018), the repeatability of the RFM has thus far been reported for rotations of the main foot joints in walking only, and no repeatability of kinematic data in running has thus far been provided.

In fact, despite the large number of skin-markers based multi-segment foot models currently available, e.g. Kidder et al., 1996; Carson et al., 2001; Leardini et al., 2007; Bishop et al., 2013, few of these have been thoroughly tested for repeatability in standard gait-analysis tasks (Kidder et al., 1996; Carson et al., 2001; Leardini et al., 2007; Bishop et al., 2013) and scarce is the current knowledge on the effects of high-energy activities, such as running, on measurements repeatability with respect to lower-energy locomotion such as walking. A larger repeatability helps increasing the statistical power and decreasing the minimal detectable difference when assessing group effects.

Physiological alterations in the execution of a motor task and errors in the methodology and instrumentation may both affect the variability of kinematic measurements (Newell 1998). In addition to the variability in motor task execution (Bartlett et al., 2007), which is independent from the measuring system, the two main sources of variability in skin-markers based kinematic measurements are due to inconsistent markers' placement (Carson et al., 2001; Caravaggi et al., 2011), by different examiners or across sessions, and the soft tissue artifacts (Tranberg and Karlsson, 1998). Walking and running are both complex multiple degree of freedom motor tasks entailing motion of the foot and lower limb joints, thus are subjected to natural variability (Davids et al., 2003; Bartlett et al., 2007).

Different methodological approaches have been proposed to get better insight into within- and between- subjects' variability (Hunter et al., 2004; Mullineaux et al., 2004; Schwartz et al., 2004). Schwartz et al. (2004) suggested that within-subject, within-observer, and between-observer errors of kinematic measurements can be identified beyond the natural variability of the motor task. However, scarce is the current understanding on how soft tissue artifacts affect skin-markers based foot kinematics in running compared to walking, therefore their effect on repeatability of these measurements is difficult to predict. In general, identifying the contribution of each source of variability in the kinematic measurement is not simple. Thus, in this study, the term "variability" will express the combination of the inherent motor task variability and the methodological sources of errors.

The main goal of this study was to assess the inter-trial, inter-session and inter-examiner repeatability of skin-markers based kinematic measurements of foot joints via the RFM in barefoot level walking and running. It was hypothesized that the repeatability of kinematic measurements would be lower in running compared to that in walking.

Methods

Two healthy subjects (subject A: female, 30 yrs, 57 kg, 1.54 m, Arch Index = 0.22, Foot Posture Index = +2; subject B: male, 26 yrs, 74 kg, 1.76 m, Arch Index = 0.26, Foot Posture Index = +3) were recruited in the study. The shank and foot were instrumented with 16 reflective skin-markers according to the RFM (Leardini et al., 2007; Portinaro et al., 2014) by four examiners in three sessions, one week apart (Schwartz et al., 2004). The RFM allows to measure rotation in the three anatomical planes between shank and foot (ShFo), shank and calcaneus (ShCa), calcaneus and midfoot (CaMi), midfoot and metatarsus (MiMe), calcaneus and metatarsus (CaMe) and first metatarsus and hallux (MeHa). Seven additional clinically meaningful angles were calculated: F2G, the sagittal-plane inclination of the 1st metatarsal bone to the ground; S2G, the sagittal-plane inclination of the 2nd metatarsal bone to the ground; V2G, the sagittal-plane inclination of the 5th metatarsal bone to the ground; S2F, the transverse-plane divergence between 1st and 2nd metatarsal bones; S2V, the transverse-plane divergence between 5th and 2nd metatarsal bones, and MLA, the medial longitudinal arch angle. Repeatability of the 24 RFM kinematic measures was assessed via the inter-trial, inter-session and inter-examiner variability in accordance with Schwartz et al. (2004). Three out of the four examiners had extensive experience with the present marker – set protocol. The fourth examiner, familiar with gait analysis methods in general, was trained on this protocol just before starting the data collection.

In each session, the participants walked and ran barefoot at self-selected speed on an instrumented treadmill (AMTI, Watertown, MA). An eight-camera motion analysis system (Vicon, Oxford, England) collected 3D kinematic data at 200 Hz. Both subjects were deemed rearfoot strikers after visual assessment of videos from high-speed cameras (125 Hz). Foot markers trajectories were filtered with a Woltring low-pass filter (cutoff frequency = 10 Hz) and processed in Visual3D (C-Motion, Germantown, MD). Joint rotations were calculated using the Joint Coordinate System (Grood and Suntay, 1983) convention. The axes of each joint reference frame were defined as follows: sagittal plane rotations around axis z (medio-lateral); frontal plane around axis x (anterior-posterior); and transverse plane rotations around axis y (vertical). Ground reaction forces were recorded at 1000 Hz for gait cycle phases' determination. Data were normalized to 0–100% of stance phase.

The offset variability across measurements of each kinematic variable was determined according to Schwartz et al. (2014). This is calculated as the average - across normalized time duration - of the frame-by-frame standard deviation across trials, which were pooled as follows: inter-trial, across 24 groups (4 examiners*2 subjects*3 sessions) of 5 trials; inter-session, across 8 groups (4 examiners*2 subjects) of 15 trials; inter-examiner, across 2 groups (2 subjects) of 60 trials for each walking and running.

According to Shapiro-Wilk test, most of the offset variability of kinematic measures was not normally distributed ($p>0.05$). Therefore, Wilcoxon signed rank test with Bonferroni-correction was used to find any significant difference in variability between walking and running Correlation analysis identified five independent variables of inter-session variability, thus an adjusted alfa = 0.01 was used when comparing intersession variability between walking and running. 1D-Statistical parametric mapping (1D-SPM) was used to assess repeatability of kinematic measurements in terms of full patterns (Pataky, 2010). This was achieved by comparing groups of 5 trials each across examiners and sessions. To assess inter-examiner repeatability, t-tests were used to perform 36 group-to-group comparisons for each kinematic measure: 6 comparisons*3 sessions*2 subjects. To assess inter-session repeatability, 24 comparisons were performed for each kinematic measure: 3 comparisons*4 examiners*2 subjects. In order to assess differences in the temporal pattern of measurements regardless of the initial offset, the joint rotations in bipedal standing posture recorded in each session for each examiner were removed from the corresponding kinematic data. According to the outcome of each group-to-group comparison, this was scored as follows: repeatable, if no statistical difference was found; largely repeatable, if the total suprathreshold cluster was less than 20% of the whole-time interval; lowly repeatable, if the total suprathreshold cluster was between 21-99% of the whole-time interval, and no repeatable if the suprathreshold cluster was equal to 100% of the time interval (see figure 2).

The study was approved by the Research and Ethics Committee of the School of Medicine, University of Sao Paulo (#031/15) and all participants gave informed consents prior to participation.

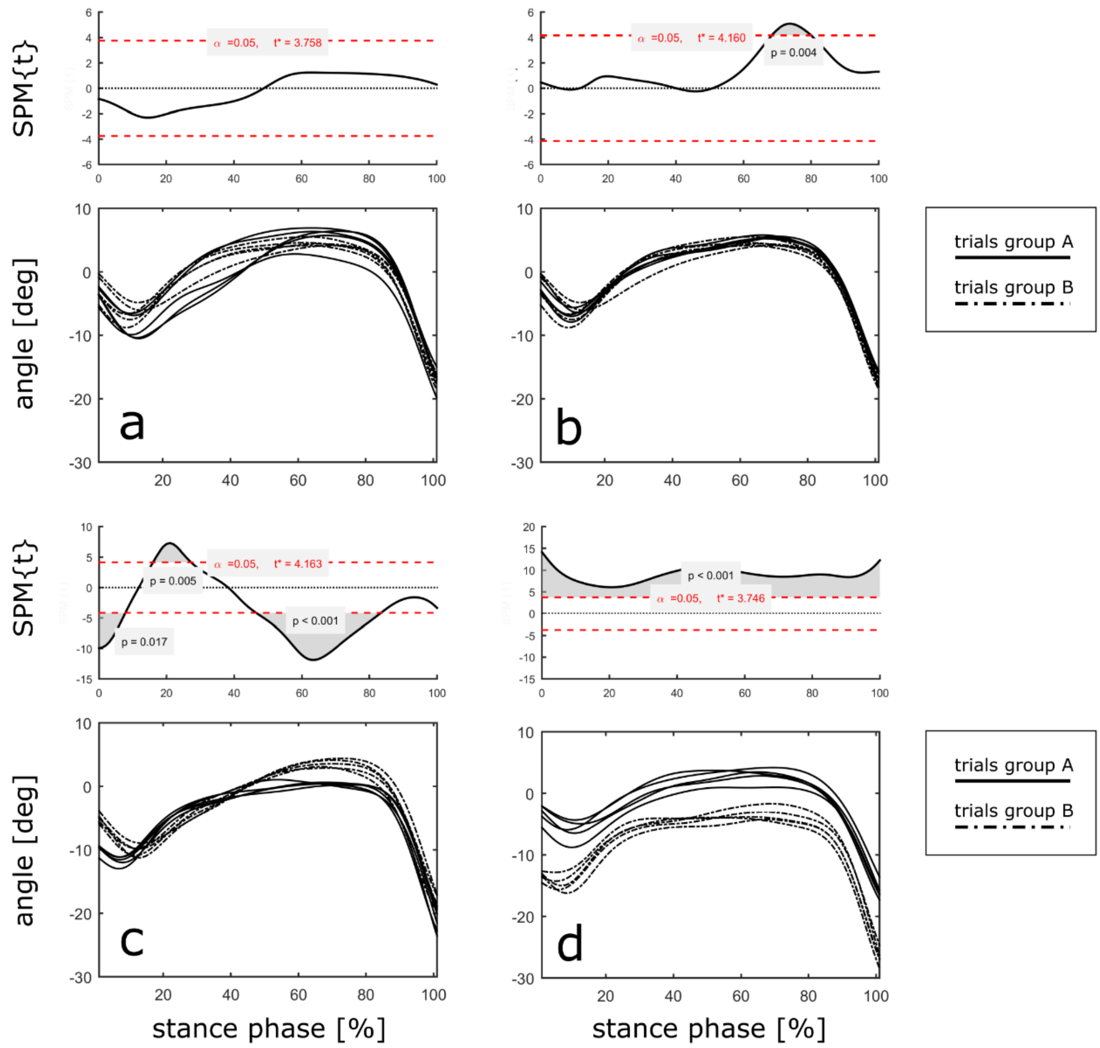

Figure 2 - Exemplary 1D-SPM comparisons between two groups of trials for sagittal plane motion between shank and calcaneus in walking. According to the outcome of the group-to-group comparison, kinematic data were classified as: (a) repeatable, when no difference was detected between the two groups of trials over stance duration; (b) largely repeatable, if the suprathreshold cluster was smaller than 20% of the stance duration; (c) lowly repeatable, if the suprathreshold cluster was between 20 and 99% of the stance duration, and (d) no repeatable, if the suprathreshold cluster was equal to 100% of the stance duration.

Results

For each kinematic measure, the inter-examiner offset variability was larger than the inter-session and the latter was larger than the inter-trial (figure 3). Respectively in walking and in running, the average inter-trial variability (deg, ±SD) across measurements was 1.0±0.3 (range 0.5–1.6) and 0.8±0.3 (0.3–1.4 deg), the inter-session was 3.9±1.4 (1.9–7.4) and 4.4±1.5 (2.1–7.3), and the inter-examiner was 5.4±2.3 (2.4–11.4) and 5.7± 2.2 (2.8–10.8). The largest inter-examiner variability was observed for sagittal-plane the calcaneus-metatarsus angle (CaMe-z) and first metatarsophalangeal joint rotations (MeHa), and the lowest for the sagittal-plane rotations of the shank-foot angle (ShFo-z) (figure 3). Wilcoxon signed rank test identified 9 kinematic measures with slightly larger inter-trial variability in walking compared to running ($p<0.05$; range difference: 0.1–0.5 deg) and 2 kinematic measures with slightly larger inter-session variability in running ($p<0.01$; range: 0.7–1.1 deg) (see Fig. 3).

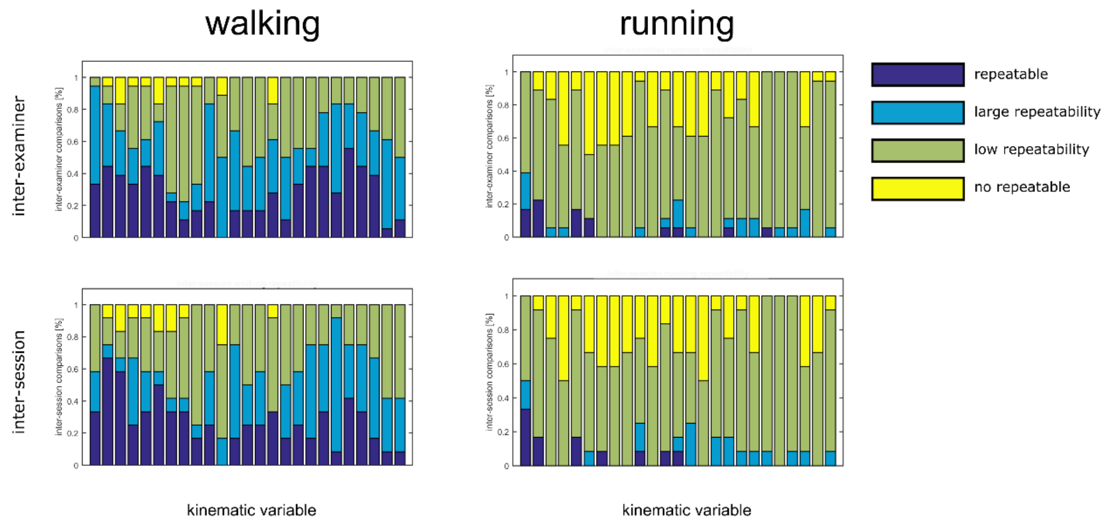

Figure 3 - For each kinematic measure, percentage of group-to-group comparisons, which were deemed as repeatable, largely repeatable, lowly repeatable and no repeatable inter-session and inter-examiner for the two motor tasks (see Fig. 1).

The outcome of the repeatability assessment via 1D-SPM in walking and running is shown in figure 2. For each kinematic variable, it is reported the percentage of group-to-group comparisons, which resulted repeatable, largely repeatable, lowly repeatable and no repeatable (Fig 4). In walking, most kinematic measures were repeatable or largely repeatable. Only motion between midfoot and calcaneus (CaMi), and transverse plane rotations between metatarsus and midfoot (MiMe-y) were, for most comparisons, lowly repeatable inter-examiner and inter-session. In running, all variables were mostly low or no repeatable (see Fig. 4).

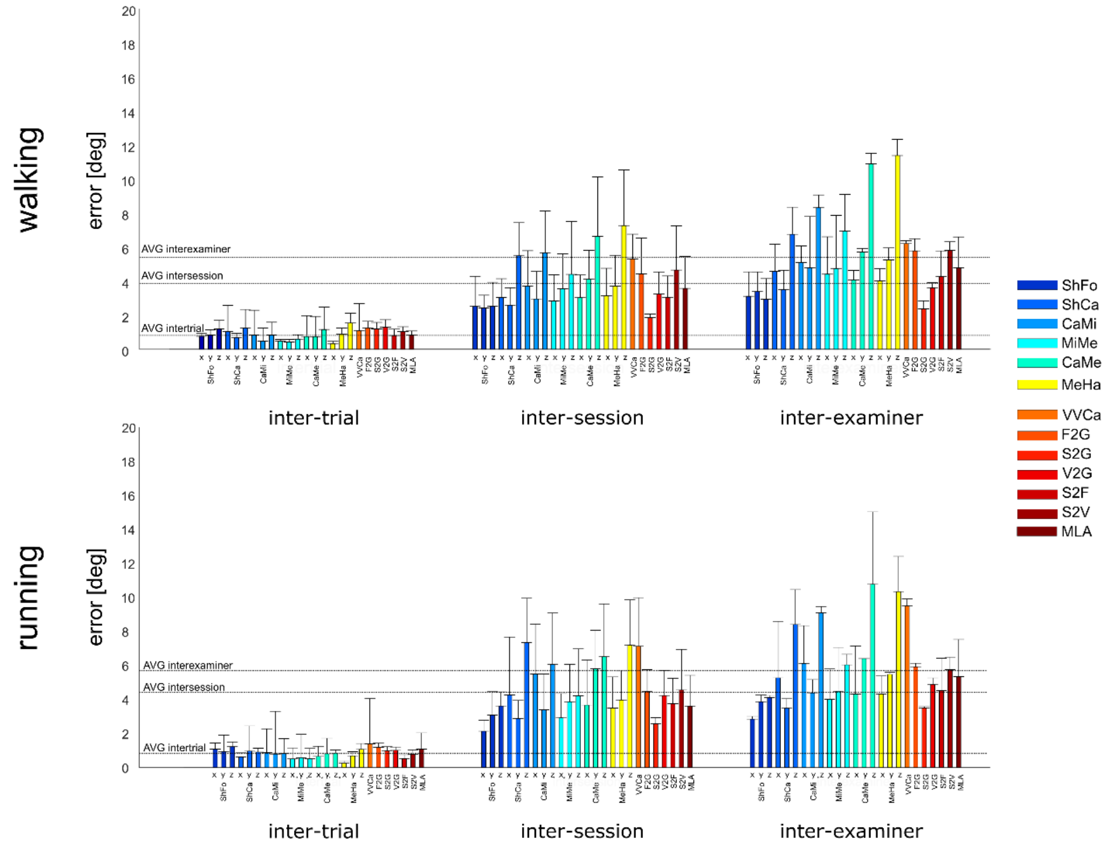

Figure 4 - Inter-trial, inter-session and inter-examiner offset variability of 24 kinematic measures from the Rizzoli Foot Model during stance phase of walking (top) and running (bottom). Average offset variability across all kinematic variables in the same variability group are shown as dotted straight line.

Discussion

Repeatability of kinematic measurements should be acknowledged or carefully assessed in order to properly design a study in terms of sample size and to allow correct interpretation of intra- and inter-subject differences. The offset variability in the main foot joint rotations and in the medial longitudinal arch deformation calculated here in walking were consistent with those reported previously using the same kinematic protocol (Caravaggi et al., 2011, 2019).

Moreover, similar to what reported before, inter-examiner variability was larger than inter-trial and inter-session, both in walking and running (Caravaggi et al., 2011, 2019).

As far as motor task effect is considered, inter-trial variability was lower for nine kinematic measures, and inter-session variability was larger for two kinematic measures in running with respect to walking. Although for most measures walking and running showed similar offset variability, the repeatability assessment of patterns via 1D-SPM analysis revealed that skin-markers based foot joint motion is highly variable across examiners and sessions. While it is difficult to tell apart the contribution of the natural motor task variability from the errors due to skin-markers placement and skin-motion artifacts on the observed low repeatability of kinematic patterns, running showed a larger variability of skin-markers based foot joints motion with respect to walking, thus confirming the hypothesis of our study. This information should be accounted for when comparing kinematic data between groups (e.g., pathological vs. healthy control) as shape differences in the patterns– such as different normalized time-points of minimum-maximum joint rotations - may not indicate kinematic alterations due to the pathology or any other variable analyzed, but could be the consequence of measurements' variability, including errors in markers' placement across sessions.

Similar to what observed in this study, there seem to be a significant examiner effect on the repeatability of some kinematic measurements, such as the S2F angle and the rotations involving calcaneus and midfoot (Caravaggi et al., 2011; Deschamps et al., 2012a). The largest variability inter-session was found for sagittal-plane rotations between shank and calcaneus, whereas the largest inter-examiner variability for the calcaneus-metatarsus joint, in both walking and running. These results are consistent with what reported by Caravaggi et al. (2011), suggesting that small differences in the position of the markers on the calcaneus could result in large variability of the frame-by-frame measurements entailing this segment. The variability in frontal-plane alignment of the calcaneus (VVCa), sagittal-plane rotation between calcaneus and metatarsus (CaMe-z) and between metatarsus and hallux (MeHa-z) were larger than 5 deg for both walking and running, thus particular attention should be paid when assessing those measures. Our findings further stress the need for experienced examiners in markers positioning especially when collecting data in different sessions.

While subjects walked and ran at their self-selected comfortable speed on a treadmill to minimize the natural motor task variability (Dingwell et al., 2001; Jordan and Newell, 2008; Wheat et al., 2005), the present analysis could not distinguish the source of variability in the measurements. As expected, natural motor task variability could be confused with experimental error. Estep et al. (2018) have reported larger natural variability in running with respect to walking, which may have contributed to the lower repeatability of treadmill running kinematic measurements observed in this study. According to Schwartz et al. (2004) the inter-trial variability could be used as an indicator of the motor task natural variability, and to assess extrinsic variability. Further studies should therefore be sought to estimate the weight of the motor task natural variability with respect to other sources of errors.

According to the results of this study, shape-similarity of kinematic patterns appear to be highly affected by the motor task, with running being less repeatable than walking. Although confirmation on a larger population and with different kinematic protocols should be sought, attention should be paid in the interpretation of skin-markers based kinematics in running across sessions or involving multiple examiners.

Conflict of Interest

The authors affirm that this study has not received any funding/assistance from a commercial organization that could lead to a conflict of interest.

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3.2 Reliability of medial-longitudinal-arch measures for skin-markers based kinematic analysis

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Short communication

Reliability of medial-longitudinal-arch measures for skin-markers based kinematic analysis

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Abstract

The medial-longitudinal arch (MLA) is perhaps the most important feature characterizing foot morphology. While current skin-markers based models of the MLA angle used in stereophotogrammetry allow to estimate foot arch shape and deformation, these do not always appear consistent with foot anatomy and with standard clinical definitions. The aim of this study was to propose novel skin-markers based measures of MLA angle and investigate their reliability during common motor tasks. Markers on the calcaneus, navicular tuberosity, first metatarsal head and base, and on the two malleoli were exploited to test eight definitions of MLA angle consistent with foot anatomy, both as angles between two 3-dimensional vectors and as corresponding projections on the sagittal plane of the foot. The inter-trial, inter-session and inter-examiner reliability of each definition was assessed in multiple walking and running trials of two volunteers, tested by four examiners in three sessions. Inter-trial variability in walking was in the range 0.7–1.2 deg, the inter-session 2.8–7.5 deg, and the inter-examiner in the range 3.7–9.3 deg across all MLA definitions. The Rizzoli Foot Model definition showed the lowest inter-session and inter-examiner variability. MLA measures presented similar vari- ability in walking and running. This study provides preliminary information on the reliability of MLA measurements based on skin- markers. According to the present study, angles between 3-dimensional vectors and minimal marker sets should be preferred over sagittal-plane projections. Further studies should be sought to investigate which definition is more accurate with respect to the real MLA deformation in different loading conditions.

Keywords: Medial longitudinal arch Skin-markers Foot kinematics Walking Running Reliability

Introduction

The Medial-longitudinal Arch (MLA) is perhaps the single most important feature used to describe foot morphology and mechanics. In addition to the weight-bearing properties in static postures, the skeletal and ligamentous structures comprising the MLA are an important elastic storage-return mechanism (Hicks, 1954; Ker et al., 1987), and have a major role in transferring and dampening forces through the foot during dynamic tasks (Caravaggi et al., 2009; Nachbauer and Nigg, 1992; Saltzman and Nawoczenski, 1995; Stearne et al., 2016). Morphology and mechanics of the MLA are multidisciplinary topics, relevant to human anthropology (Bennett et al., 2009; Griffin et al., 2015), to sport and footwear biomechanics (Lin et al., 2012; Perl et al., 2012), and to foot pathologies associated to alterations of MLA shape (Pfeiffer et al., 2006; Tome et al., 2006). In postural control, decreased mobility of the MLA was found to increase sway even after small perturbations (Birinci and Demirbas, 2017) further emphasizing the importance of accurate measurement of MLA dynamics and its relationships with foot functionality.

Although no consensus has been reached in the literature, the most widely-used clinical measures to quantify objectively MLA posture and deformation are the arch height index (Williams and McClay, 2000), the relative arch deformation (Nigg et al., 1998), and the navicular drop index. (Saltzman et al., 1995). The latter, still rather common in the clinical practice, has been shown to be correlated with rearfoot pronation (Boozer et al., 2002), which is suggested to be associated to the stability of the MLA. In addition, footprint-based parameters such as the arch index (Cavanagh and Rodgers, 1987), the arch-length index (Hawes et al., 1992) and the Staheli's index (Staheli et al., 1987) have also been devised. In particular, it has been shown that the arch index can explain 50% of variance in MLA height (McCrory et al., 1997). Scores based on visual observation and manual palpation, such as the Foot Posture Index, have also been proposed (Redmond et al., 2006), but this is associated more with the foot pronation/supination posture which is not strictly related to MLA shape.

Stereophotogrammetry, in combination with multisegment foot models, allow to replicate the aforementioned clinical definitions also in dynamic tasks (Hunt et al., 2000; Jenkyn and Nicol, 2007) and to track the MLA angle over time. Most of current geometrical approximations of the MLA angle reported in multisegment foot protocols entail the use of anatomical landmarks on the calcaneus, on the navicular bone and on the first metatarsal head (Bandholm et al., 2008; Simon et al., 2006; Tome et al., 2006). MLA posture is generally calculated as the angle between two 3-dimensional vectors bounded by those markers, with no particular assumptions on their orientation with respect to foot anatomy. In an effort to better replicate the anatomical shape of the MLA, a parabola (Perl et al., 2012) and an ellipse (Ikeda et al., 2014), which best interpolate the position of those skin-markers, have also been proposed. In order to improve the consistency with traditional clinical measures of MLA based on lateral x-ray images, the Rizzoli Foot Model (RFM) is currently the only multisegment foot model approximating the MLA arch as the angle between the projections on the sagittal plane of the foot of two line segments connecting at the sustentaculum tali (Leardini et al., 2007; Portinaro et al., 2014).

While the importance for measuring the MLA is generally recognized, no consensus has thus far been reached on which MLA model better represents the foot medial longitudinal arch shape and deformation. Some of the current definitions used in gait analysis, based on skin-markers, do not appear consistent with foot anatomy and mechanics. Furthermore, despite the importance of MLA biomechanics across several disciplines, a thorough investigation of the error in the calculation of MLA deformation during common motor tasks has yet to be performed. The aim of this study was to compare eight skin-markers based definitions of MLA angle, both as angles between 3-dimensional vectors and as angles between corresponding projections on the sagittal plane of the foot, in terms of inter-trial, inter- and intra-examiner reliability during walking and running.

Methods

Skin-markers based definitions of MLA

The eight geometrical definitions of MLA were based on skin-markers located on the calcaneus (CA: upper central ridge of the calcaneus posterior surface), sustentaculum tali (ST: most medial apex), talo-navicular tuberosity (TN: most medial apex of the navicular tuberosity), the two malleoli (distal apex of the medial and lateral malleolus), base of the first metatarsal bone (FMB: dorso-medial aspect of the first metatarso-cuneiform joint) and head of the first metatarsal bone (FMH: dorso-medial aspect of the first metatarso-phalangeal joint) (Figure 5), as described in (Leardini et al., 2007; Portinaro et al., 2014). These MLA definitions aim at mimicking the Moreau-Costa-Bertani angle (Moreau and Costa-Bertani, 1943) and the angle between rearfoot inclination and first metatarsal inclination described by Saltzman et al. (1995). All MLA measurements were calculated: 1) as angles between two 3-dimensional vectors (here called 3D angles), and 2) as angles between the projections of those vectors on the sagittal plane of the foot (here called 2D angles, α in equation 1). This is defined as the plane orthogonal to the transverse plane (through CA, FMH and head of the fifth metatarsal bone), and passing through CA and head of the second metatarsal bone (Cappozzo et al., 1995).

$$\alpha = \cos^{-1}(\vec{u} \cdot \vec{v}) \quad \text{Equation 1}$$

where *u* and *v* are two-unit vectors, the directions of which are established by real markers on anatomical landmarks or by “virtual” markers constructed from a set of real and virtual points.

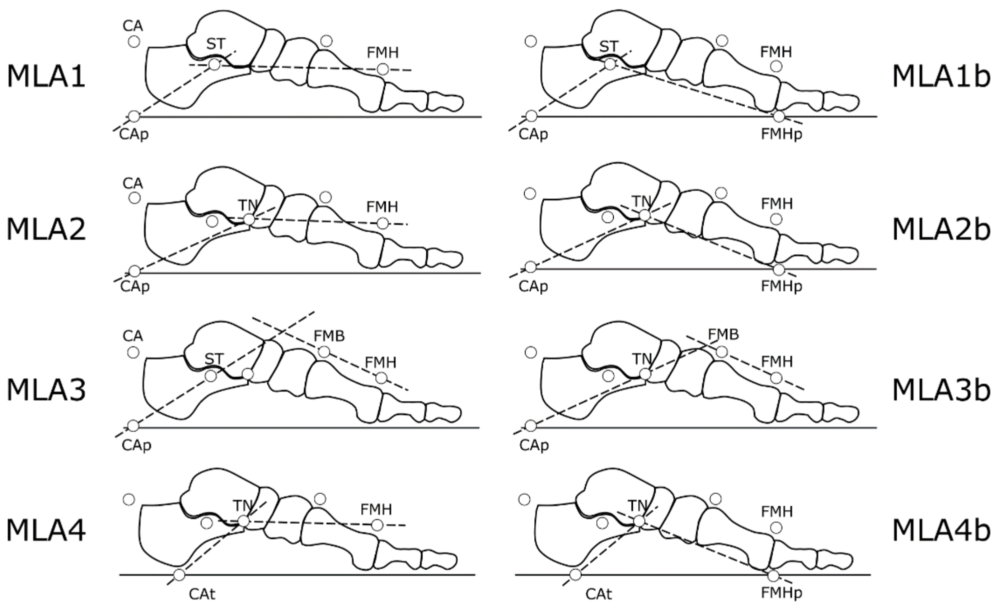

Figure 5 - Left (top to bottom), four skin-markers based measures of medial longitudinal arch angle. Right, four variations using either the projection of FMH on the ground (MLA1b, MLA2b and MLA4b) and using TN instead of ST (MLA3b).

A detailed description of each marker-based MLA measure follows (see figure 5).

- **MLA1 - or RFM definition:** the vector on the proximal segment is bounded by marker CA_p and ST, where CA_p is the projection of CA on the x-y plane of the laboratory reference frame (i.e., the ground). The vector on the distal segment is bounded by markers FMH and ST.
- **MLA1-b:** as MLA1 by replacing FMH with FMH_p, where FMH_p is the projection of FMH on the ground.
- **MLA2 - or talo-navicular apex definition:** the vector on the proximal segment is bounded by marker CA_p and by TN. The vector on the distal segment by FMH and TN.
- **MLA2-b:** as MLA2 by replacing FMH with FMH_p.

- **MLA3 - or first metatarsal bone orientation definition:** the vector on the proximal segment is bounded by markers CAp and ST. The vector on the distal segment by FMH and FMB.
- **MLA3-b:** as MLA3 by replacing ST with TN.
- **MLA4 - or calcaneal tuberosity definition:** the vector on the proximal segment is bounded by CA_t and TN, where CA_t is the projection of CA_m - midpoint between CA and the midpoint between peroneal tubercle and ST - on the ground plane. The vector on the distal segment is bounded by FMH and TN.
- **MLA4-b:** as MLA4 by replacing FMH with FMH_p

Each of the above MLA angles was calculated both as the 3D and 2D angles, for a total of 16 MLA measures.

The virtual markers' positions (such as CA_p) were established with the subject in static bipedal standing posture. During kinematic analysis, the trajectory of virtual markers was rigidly fixed to that of the relevant segment local reference frame.

Repeatability study

A repeatability study was performed to evaluate the inter-trial, inter- and intra-examiner error in calculating MLA angle according to the 16 definitions. The study entailed measuring MLA angle during static bipedal standing posture, and temporal profiles during walking and running in two subjects (subject A: female; 30 years; 57 kg; 1.54 m; Arch Index = 0.22; Foot Posture Index = +2; subject B: male; 26 years; 74 kg; 1.76 m; Arch Index = 0.26; Foot Posture Index = +3). The shank and foot of these subjects were instrumented with reflective skin-markers by four examiners in three sessions, one week apart (Schwartz et al., 2004). An 8-camera motion analysis system (Vero Vicon; sampling rate = 100 Hz) was used to track 16 markers, according to the RFM (Leardini et al., 2007; Portinaro et al., 2014). Markers trajectories were filtered with a Woltring low-pass filter (cutoff frequency = 10 Hz) and imported as .c3d files in Visual3D (Visual3D, C-Motion, Germantown, MD) for angles calculation. The variability of MLA measures was computed as the average of the standard deviation over the gait cycle (AVG-SD), across a number of walking trials pooled according to Schwartz et al. (2004). Accordingly, the variability was determined from the average of this error as follows: inter-trial, across 24 groups (4 examiners*2 subjects*3 sessions) of 5 trials; inter-session, from 8 groups (4 observers * 2 subjects) of 15 trials; inter-examiner, from 2 groups (2 subjects) of 60 trials.

Results

Mean temporal profiles of MLA angles across 60 walking and 60 running trials for one of the two subjects, along with mean MLA angles during static bipedal standing posture, are shown in figure 6 and figure 7, respectively. In general, the 2D angles were larger than the corresponding 3D angles.

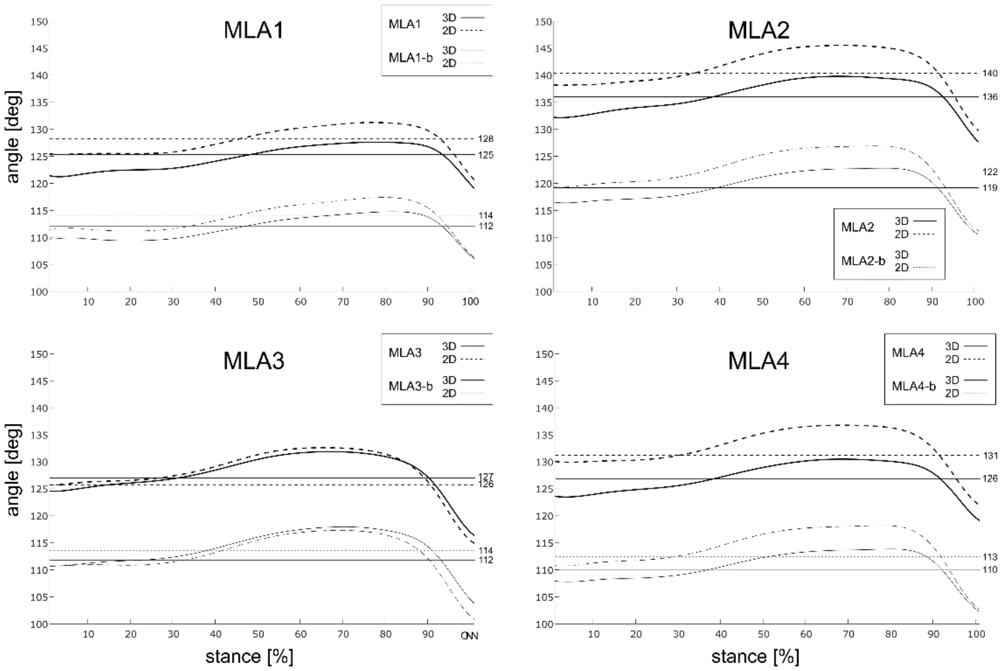

Figure 6 - Mean temporal profiles of MLA angles during normalized stance phase duration for one of the two subjects across 60 walking trials. MLA angles are shown as continuous bold line (3D angle), dotted bold line (2D angle), continuous thin line (3D angle, b variation) and dotted thin line (2D angle, *b* variation). For each definition, straight lines are showing the mean MLA angle in bipedal standing posture. See figure 5 for measures description.

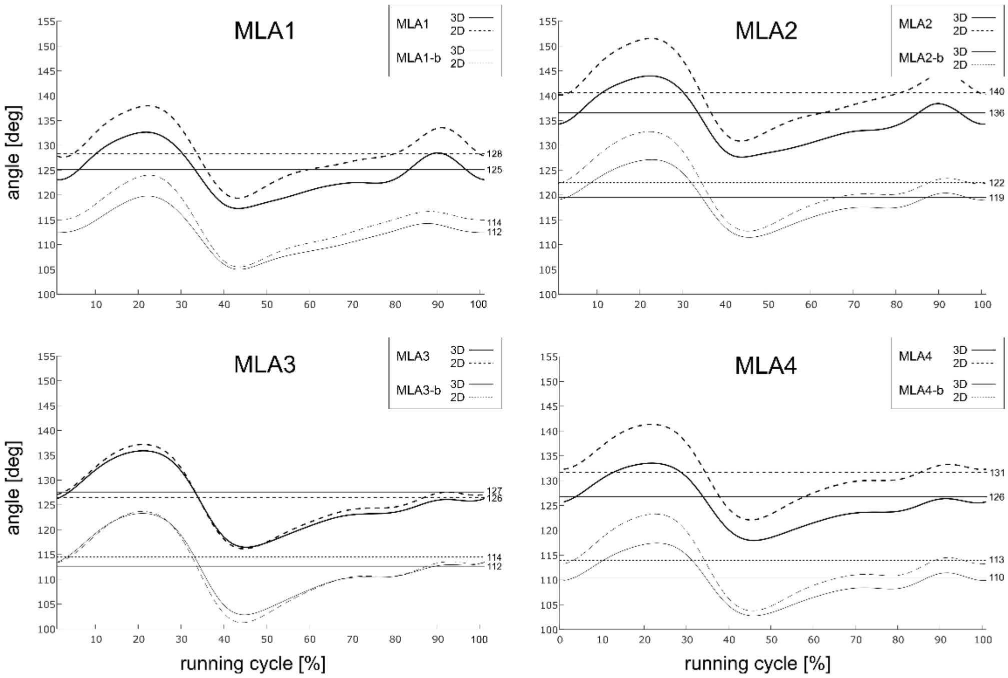

Figure 7 - Mean temporal profiles of MLA angles during normalized running cycle for one of the two subjects across 60 walking trials. MLA angles are shown as continuous bold line (3D angle), dotted bold line (2D angle), continuous thin line (3D angle, b variation). For each definition, straight lines are showing the mean MLA angle in bipedal standing posture. See figure 5 for measures description.

The inter-trial variability in walking was in the range 0.7 – 1.2 deg, the inter-session 2.9 – 7.7 deg, and the inter-examiner in the range 3.7 – 9.3 deg across all MLA definitions (table 1). MLA1, and its variation MLA1-b, showed the lowest inter-session and inter-examiner variability in both walking and running. MLA measures showed similar variability in walking and running. For each MLA definition, the variability of the 3D angles was always smaller than the corresponding 2D angles.

Table 1 - Inter-trial, inter-session and inter-examiner variability [deg] of MLA measures in walking and running. Variability data are reported for each of the eight measures calculated both as angles between 3-dimensional vectors (3D), and as angles b between corresponding projections on the sagittal plane of the foot (2D). See figure 5 for details.

		Variability - AVG-SD [deg]					
		inter-trial		**inter-session**		**inter-examiner**	
		walking	running	walking	running	walking	Running
MLA1	**3D**	0.7	0.7	2.8	3.0	3.7	4.0
	2D	0.8	1.0	3.6	3.3	4.8	4.6
MLA1-b	**3D**	0.7	0.7	3.7	3.5	5.1	5.2
	2D	0.8	0.9	4.3	4.0	6.0	5.8
MLA2	**3D**	0.8	0.7	3.4	3.3	4.2	4.2
	2D	1.1	1.0	5.2	4.9	6.9	6.6
MLA2-b	**3D**	0.8	0.7	4.5	4.3	5.9	6.1
	2D	1.0	0.9	5.7	5.1	7.7	7.6
MLA3	**3D**	0.9	0.8	5.4	4.2	6.5	5.4
	2D	1.2	1.1	7.5	6.2	9.3	7.9
MLA3-b	**3D**	0.9	0.8	4.2	3.8	5.7	5.1
	2D	1.1	1.1	6.2	5.5	8.2	7.3
MLA4	**3D**	0.7	0.7	3.9	3.4	4.8	4.4
	2D	1.1	1.0	5.9	5.4	8.1	7.3
MLA4-b	**3D**	0.7	0.7	4.9	4.5	6.6	6.5
	2D	1.1	1.0	6.4	5.8	8.9	8.5

In walking, the inter-trial (n = 60) median range of motion (ROM) ranged between 11 – 18 deg for subject 1 and 10 – 17 deg for subject 2, across the MLA definitions (table 2). In running, ROM ranged between 15 – 22 deg for subject 1, and between 14 – 28 deg for subject 2. MLA1 showed the lowest ROM in both walking and running. For each MLA definition, ROM of the 3D angles in walking and running was lower than that for the corresponding 2D angles.

Table 2 - For each MLA definition, the inter-trial median [25% - 75%] ROM [deg] in walking and running for the two subjects. ROM data are reported for each of the 8 measures calculated both as angles between 3-dimensional vectors (3D), and as angles between corresponding projections on the sagittal plane of the foot (2D). See figure 5 for details.

MLA definition		ROM [deg]			
		Walking		Running	
		Subject 1	*Subject 2*	*Subject 1*	*Subject 2*
MLA1	**3D**	9 [8 11]	7 [6 7]	16 [15 17]	15 [12 19]
	2D	11 [10 12]	10 [9 11]	19 [18 21]	22 [20 26]
MLA1-b	**3D**	9 [8 10]	7 [6 7]	15 [14 16]	15 [12 19]
	2D	12 [10 13]	10 [9 11]	19 [18 20]	21 [19 26]
MLA2	**3D**	13 [11 15]	9 [9 11]	17 [16 19]	17 [14 20]
	2D	16 [15 18]	16 [15 18]	21 [20 24]	28 [25 30]
MLA2-b	**3D**	13 [11 14]	9 [10 12]	17 [15 17]	17 [14 18]
	2D	16 [15 17]	16 [14 17]	20 [19 22]	24 [22 26]
MLA3	**3D**	17 [15 18]	14 [12 16]	20 [19 21]	19 [18 24]
	2D	18 [17 20]	17 [16 20]	21 [20 24]	26 [25 31]
MLA3-b	**3D**	15 [13 17]	12 [10 14]	21 [20 23]	21 [17 24]
	2D	17 [15 19]	16 [13 18]	22 [21 25]	27 [25 31]
MLA4	**3D**	12 [10 13]	10 [9 11]	16 [15 18]	14 [13 18]
	2D	15 [14 17]	15 [14 18]	20 [18 22]	24 [22 27]
MLA4-b	**3D**	12 [11 13]	11 [10 12]	15 [14 16]	16 [14 18]
	2D	16 [15 17]	16 [15 17]	20 [18 22]	24 [22 26]

Discussion

The main objective of this study was to establish a set of possible geometrical definitions of the MLA angle based on skin-markers, and to assess their repeatability during common locomotor activities with respect to traditional definitions. In order to be consistent with clinical/radiological measures, these were calculated both as angles between 3-dimensional vectors and as angles between the projected vectors on the sagittal plane of the foot. For each MLA definition, variability and ROM in walking and running of the 3D angles were always lower than the corresponding 2D angles. This could be accounted for to the additional error in the definition of the sagittal plane of the foot, on which markers' positions are projected. For the same reason, larger variability was also detected for MLA definitions based on larger number of real and virtual markers, such as MLA3 and MLA4. As expected, the inter-examiner variability was larger than the inter-session and this was larger than the inter-trial across all MLA measurements. Despite the larger accelerations the foot is subjected to in running, no differences in variability were detected with respect to the corresponding errors in walking. While the present inter-trial and inter-session variability of MLA measurements is rather consistent with those previously reported, the average inter-examiner repeatability is slightly lower than what calculated for other foot joints during walking (range 2.7 – 11.5 deg, from Caravaggi et al. (2011)). This result is remarkable considering the larger angles and overall motion measured with the present MLA definitions, and further stresses the need for experienced operators in markers positioning as those recruited in the present investigation.

This study allows quantifying the systematic errors in MLA measurements that should be accounted for when assessing differences between groups involving more than one observer in the data collection. The present results may help choosing the measure with higher reliability which will give greater statistical power to detect differences between groups, for a given sample size. While knowledge of the inter-trial and inter-examiner reliability of MLA measures based on skin-markers allows assessing the robustness of the measurements, no information can be inferred on the accuracy of the real MLA posture and deformation in dynamic activities. The rather small variability and range of motion detected for MLA1 does not necessarily imply that the model is well replicating the mechanics of the medial arch. Therefore, while the present study is indicating that MLA angle definitions based on minimal marker sets should be pursued to improve reliability of measurements, further analysis on the accuracy of current MLA definitions with respect to standard imaging-based measures are necessary and should be sought in future investigations.

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Conflict of interests

This study has not received any funding/assistance from commercial organizations that could lead to a conflict of interest.

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3.3 Accuracy and correlation between skin-marker based and radiographic measurements of medial longitudinal arch deformation

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Short communication

Accuracy and correlation between skin-marker based and radiographic measurements of medial longitudinal arch deformation

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Abstract

Static and dynamic measurements of the medial longitudinal arch (MLA) in the foot are critical across different clinical and biomechanical research fields. While MLA deformation can be estimated using skin-markers for gait analysis, the current understanding of the correlates between skin-marker based models and radiographic measures of the MLA is limited. This study aimed at assessing the correlation and accuracy of skin-marker based measures of MLA deformation with respect to standard clinical X-ray-based measures, used as reference. 20 asymptomatic subjects without morphological alterations of the foot volunteered in the study. A lateral X-ray of the right foot of each subject was taken in monopodalic upright posture with and without a metatarsophalangeal-joint dorsiflexing wedge. MLA angle was estimated in the two-foot postures and during gait using 16 skin-marker based models, which were established according to the marker set of a validated multi-segment foot kinematic protocol. The error of each model in tracking MLA deformation was assessed and correlated with respect to standard radiographic measurements. Estimation of MLA deformation was highly affected by the skin-marker models. Skin-marker models using the marker on the navicular tuberosity as apex of the MLA angle showed the smallest errors (about 2 deg) and the largest correlations ($R=0.64$-0.65; $p<0.05$) with respect to the radiographic measurements. According to the outcome of this study, skin-marker based definitions of the MLA angle using the navicular tuberosity as apex of the arch may provide a more accurate estimation of MLA deformation with respect to that from radiographic measures.

Keywords: foot; medial longitudinal arch; skin-markers; accuracy; X-ray images.

Introduction

The arch-shaped human foot evolved from the highly deformable arboreal apes’ foot (Gebo, 1992) to the present semi-rigid structure capable of adapting to different terrains and absorbing impact forces in dynamic activities such as walking (Wang and Crompton, 2004). This shape, commonly referred to as Medial Longitudinal Arch (MLA), is comprised of the joints in the medial aspect of foot, such as the talo-navicular and the navicular-cuneiform joints. The MLA is supported by the intrinsic and extrinsic foot muscles, by strong ligaments such as the calcaneo-navicular ligament, and by the plantar aponeurosis. The MLA allows the foot to sustain body weight and to act as a spring, storing elastic energy that can be recovered during dynamic tasks (Caravaggi et al., 2009; Hicks, 1954; Stearne et al., 2016).

Alterations of the MLA structure and function are responsible for foot deformities (Franco, 1987), which are often associated with abnormal plantar pressure distribution and pain (Burns et al., 2005; Kaufman et al., 1999; Ledoux and Hillstrom, 2002; Menz et al., 2013; Song et al., 1996; Williams et al., 2001). Accurate measurement of the shape of the MLA is critical for the clinical evaluation of foot postural alterations and ailments, for foot type classification and in the design of custom orthotics and footwear (Bus et al., 2013).

However, no well-established standard method to dynamically assess MLA posture is available. Typically, most measures are based on the height of the dorsal aspect of the foot or of the navicular bone (Nigg et al., 1998; Saltzman et al., 1995; Williams and McClay, 2000), or on the footprint shape (Cavanagh and Rodgers, 1987; Hawes et al., 1992; Staheli et al., 1987; Xiong et al., 2010). While some of these measures are used in clinical practice and in biomechanics, they are limited to the assessment of MLA posture in static conditions only. Because the MLA plays an important role in the dynamic and mechanics of gait, interest has grown in the study of MLA deformation during dynamic tasks. An accurate estimation of MLA kinematics can be obtained via biplanar fluoroscopy (Balsdon et al., 2019, 2016; Wearing et al., 1998), but X-ray poses risks to patients from exposure to ionizing radiation. Skin-marker based gait analysis is the current gold standard for radiation-free measurement of lower limb kinematics (Benedetti et al., 1998; Carson et al., 2001), enabling estimation of the temporal pattern of MLA deformation in different motor tasks (Bandholm et al., 2008; Bencke et al., 2012).

In general, the MLA is defined as the angle between two vectors established according to the position of three skin-markers placed on the calcaneus, on the head of the first metatarsal and on the navicular bone (Bandholm et al., 2008; Simon et al., 2006; Tome et al., 2006). According to the Rizzoli Foot Model (RFM), the MLA can be defined as the angle between the projection, on the sagittal plane of the foot, of two vectors passing through the sustentaculum tali (Leardini et al., 2007; Portinaro et al., 2014). This definition was adopted to better replicate radiography-based clinical measures of the MLA, such as the Moreau-Costa-Bertani (MCB) angle (Carrara et al., 2020; Moreau and Costa-Bertani, 1943). However, similarly to other skin-marker based kinematic measures, MLA measurements are affected by skin-motion artifacts (Shultz et al., 2011). While some efforts have been made to determine the repeatability of RFM parameters and MLA measures in dynamic activities (Caravaggi et al., 2019; Matias et al., 2020), and to estimate the sensitivity of skin-marker measures to skin artifacts (Schallig et al., 2021), no error in MLA measurements has thus far been reported. The aim of the present study was to assess the accuracy and the correlation between different skin-marker based and radiographic measures of MLA deformation.

Material and Methods

The shank and foot of twenty healthy subjects (8 M, 12 F; 29.0 ± 8.4 years, 63.0 ± 12.6 kg, 1.70 ± 0.07 m) without lower limb pathologies or any major foot postural or morphological alteration, were outfitted with 16 reflective skin-markers according to the RFM (Leardini et al., 2007; Portinaro et al., 2014) by an experienced examiner. Measurements of MLA angle were obtained according to the eight skin-marker based models previously reported (Figure 8) (Caravaggi et al., 2019). Three main models (MLA1, MLA2 and MLA4; Figure 8, left) were established to replicate the radiography based MCB angle. In order to limit the effect of skin-motion artifacts on the vertical position of the marker on the metatarsal head, three b-variations (MLA1b, MLA2b and MLA4b; Figure 8, right) were obtained by using the projection of this marker on the ground. MLA3 was established to replicate the radiography-based Calcaneal-1st Metatarsal angle (C1MA) (Saltzman et al., 1995), and its b-variation (MLA3b) was obtained by using the marker on the navicular tuberosity instead of the one on the sustentaculum tali. For each model, the MLA was calculated as the angle between two three- dimensional vectors (3D) and as the angle between the projections of the same vectors on the sagittal plane of the foot (2D). The 2D definitions were established to better replicate planar radiography-based measures of the MLA.

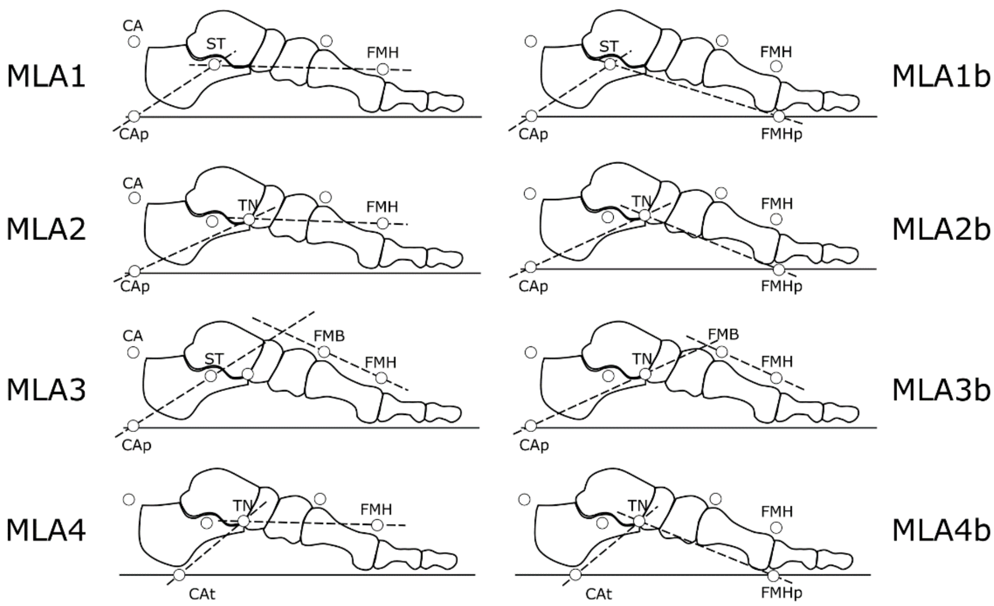

Figure 8 - The eight skin-marker based models of MLA angles used in the study. Variations MLA1b, MLA2b and MLA4b are obtained by using the projection of FMH on the ground, and MLA3b by using the marker on the navicular tuberosity (TN) instead of that on the sust the sustentaculum tali (ST). Figure taken from Figure 5 in (Caravaggi et al., 2019).

An 8-camera motion analysis system (Vicon 612, Vicon Motion Capture, Oxford, UK) was used to track the markers position in static monopodalic weightbearing standing posture with and without a toes-dorsiflexing wedge (Figure 9). A wedge with inclination of 45, 60, or 75 deg was chosen for each subject in order to achieve the maximum physiological toes dorsiflexion. The wedge helped to dorsiflex the toes and thus apply tension in the plantar aponeurosis and increase the MLA height (i.e. windlass mechanism, (Hicks, 1954)). MLA deformation was defined as the angular difference between the two-foot postures.

Accuracy of the skin-marker based measurements of MLA deformation was assessed by comparison with radiographic measurements via the MCB and the C1MA angles. The MCB is here defined as the angle, in the sagittal plane, bounded by three bony landmarks: the most inferior aspect of the talo-navicular joint (apex); the inferior border of the posterior calcaneal tuberosity, and the most inferior aspect of the head of the first metatarsal bone (Figure 8) (Carrara et al., 2020; Moreau and Costa-Bertani, 1943). The C1MA is the angle, in the sagittal plane, formed by the inferior surface of the calcaneus and a line segment parallel to the dorsum of the mid-shaft of the first metatarsal (Figure 9). A Cone Beam CT machine (Onsight 3d Extremity System, Carestream, US.) was used to acquire lateral X-rays of the feet of each subject. The X-rays were taken with the lowest radiation dose allowed by the device, with the feet in the same two postures recorded by the motion analysis system. X-rays were exported as DICOM images and processed with MicroDicom (www.microdicom.com) to measure the MCB and the C1MA angles. Only right foot data (n=20) were used in this study. Informed consent was obtained from all subjects after extensive explanation of the study aims and analysis involved.

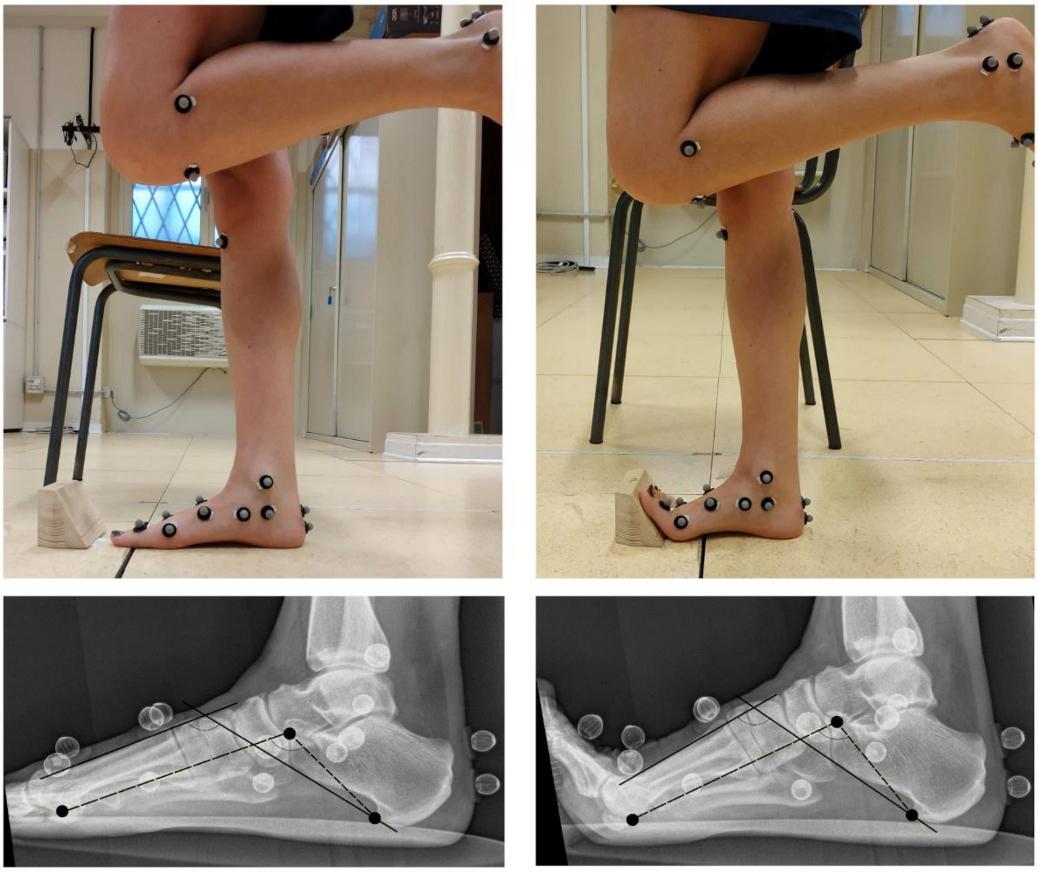

Figure 9 - Top, one of the subjects who volunteered in the study in upright monopodalic standing posture with (left) and without (right) the toes-dorsiflexing wedge. 16 skin-markers are attached to the foot and leg according to the Rizzoli Foot Model (Leardini e (Leardini et al., 2007; Portinaro et al., 2014). Bottom, corresponding sagittal-plane X-ray images of the foot and radiography-based measurements of MCB(dashed lines) and C1MA (continuous lines) angles.

Correlations between X-ray and skin-marker based measurements of MLA deformation were assessed via Pearson linear correlation. The error of each skin-marker based model in estimating MLA deformation was calculated as the absolute difference with respect to that estimated from the two X-ray measurements used as reference (see Equation 1).

$$error = |\Delta(MLASkin - markers) - \Delta(MLARX)| \text{ (Equation 1)}$$

where ΔMLASkin-markers are the MLA deformations (deg) according to the skin-marker models, and ΔMLARX the MLA deformations based on the radiographic measurements. The MCB angle was used as the reference radiographic measure for all MLA models but MLA3 and MLA3b, which were established to replicate the C1MA angle definition (Caravaggi et al. 2019). Repeatability of the two radiographic measurements was assessed via intraclass correlation coefficient ICC (2,1) on ten subjects analyzed by two examiners in different sessions using the same protocol. Bland-Altman plots were used to assess the extent and the direction of the difference between skin-marker and radiographic measurements of MLA deformation.

In order to estimate MLA deformation in a common daily motor task, markers' trajectories were also collected during three walking trials at self-selected comfortable walking speed. The range of motion (ROM) and the mean angle over gait cycle were computed for each MLA model. Static and dynamic trials were processed in Visual3D (Visual3d, C-Motion, US).

Paired non-parametric Friedman and Tukey-Kramer post-hoc tests were used to assess statistical differences in error between MLA models ($\alpha = 0.05$). A paired non-parametric Wilcoxon signed-rank test was used to analyze the differences in error between the main MLA measures and the corresponding b-variations and thus assess the effect of projecting the metatarsal bone marker. When required, Bonferroni correction was applied to the coefficient of significance to account for multiple comparisons between independent groups.

Results

Inter-subject MLA deformation was 8 ± 3 deg and 11 ± 3 deg respectively for C1MA and MCB angles. A correlation of 0.62 ($p < 0.05$) and a mean difference of 3.7 ± 2.2 deg was observed between the two radiographic measures. MLA deformations, according to models MLA2, MLA2b, MLA4 and MLA4b showed significant positive correlations with the X-ray based measurements (Table 1). All other correlations were not statistically significant ($p>0.05$). Model MLA1 showed a negative correlation, albeit not statistically significant, to the radiographic measurements.

Table 3 - For each MLA model, Pearson coefficients of correlations between X-ray based and skin-marker based measurements of MLA deformation [deg]. * is showing statistically significant correlations ($p < 0.05$).

MLA model		**Pearson Rho**
MLA1	2D	-0.29
	3D	0.25
MLA2	2D	0.64*
	3D	0.57*
MLA3	2D	0.24
	3D	0.16
MLA4	2D	0.66*
	3D	0.56*
MLA1b	2D	0.43
	3D	0.22
MLA2b	2D	0.64*
	3D	0.55*
MLA3b	2D	0.19
	3D	0.22
MLA4b	2D	0.65*
	3D	0.55*

All MLA models underestimated the radiography-based MLA deformation (Figure 11). The average absolute errors with respect to the X-ray measurements ranged between 2-10 deg, or 10 - 70% of the nominal radiographic measurements, across all models (Figures 10 & 11). The lowest errors (about 2 deg) were observed for models MLA2b and MLA4b as sagittal-plane projections, with the corresponding 3D angles being slightly larger. The largest errors (about 10 deg) were observed for models MLA1 and MLA1b. All b-variations showed lower errors than the corresponding MLA models ($p < 0.01$).

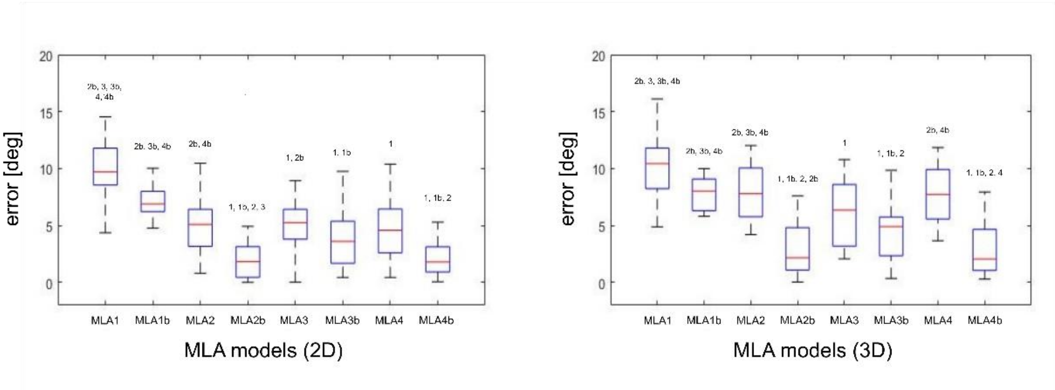

Figure 10 - For each skin-marker based MLA model, box-plot of the error distribution in measuring MLA deformation with respect to radiographic measurements. Left, errors of the 2D MLA models (angle between the projection of the relevant vectors on the sagittal plane of the foot) and, right, errors of the 3D MLA models (angle between three-dimensional vectors). The statistically significant differences between error groups are showed over each box-plot according to the paired non- parametric Friedman and Tukey-Kramer post-hoc tests.

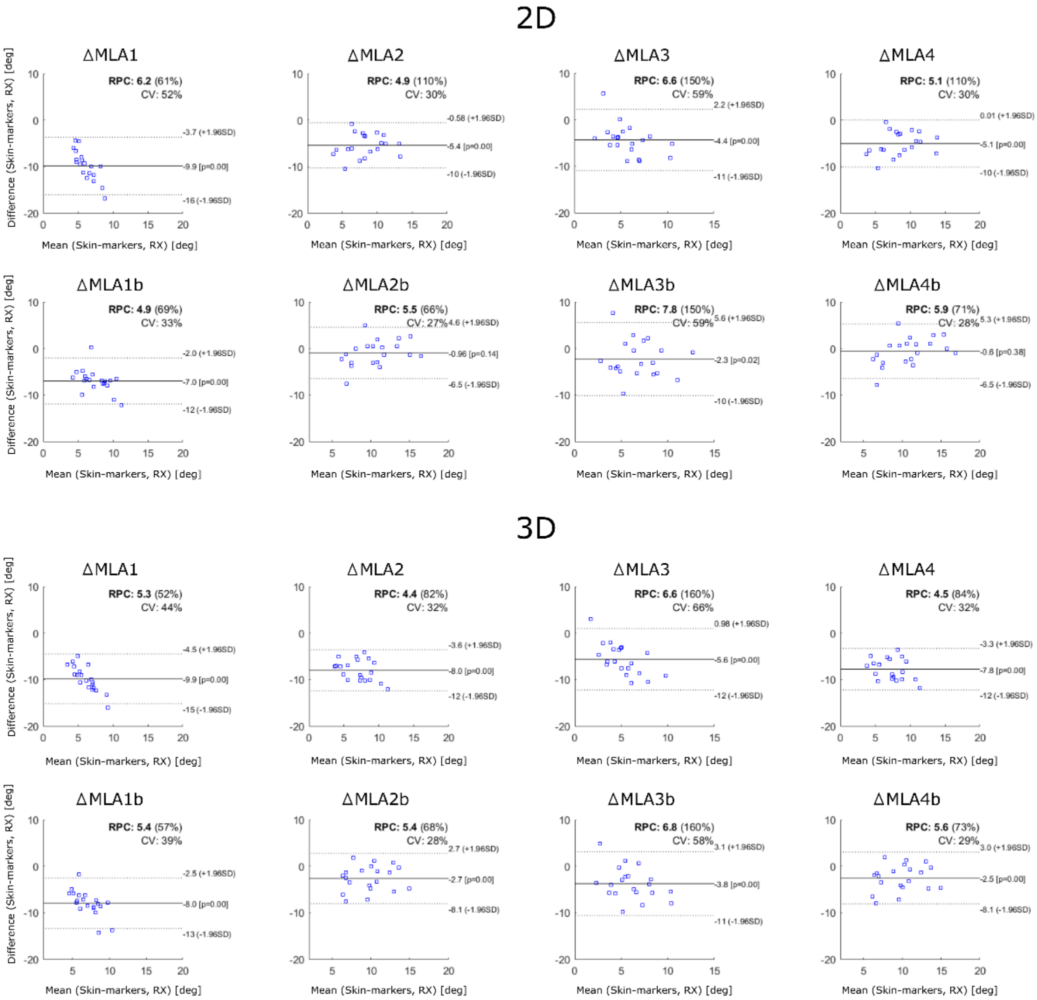

Figure 11 - For each skin-marker based MLA model, Bland-Altman plots of the agreement between skin-marker and radiographic measurements of MLA deformation (ΔMLA).

An ICC of 0.88 and 0.97 was observed for the measurements of MLA deformation according to the MCB and the C1MA angles, respectively. The inter-session errors of skin-marker based MLA measurements during upright static posture ranged between 3-5 deg (3-4 % of static MLA measurements) and were very similar across all MLA models (see also Caravaggi et al. 2019).

In the stance phase of walking, the mean inter-subject ROM of MLA angle ranged between 11-21 deg across all models (Table 2). 3D MLA definitions showed a mean ROM consistently lower (10- 17 deg) than that of the corresponding 2D definitions (16-21 deg). The mean MLA angles in gait ranged between 115 – 145 deg across all models.

Table 4 - For each skin-marker based MLA model, inter-subject (n=20) mean (±std) ROM during stance phase and gait cycle. The mean MLA angle over gait cycle is reported in the last column.

MLA model		ROM [deg] *stance*	ROM [deg] *stride*	Mean angle [deg] *Stride*
MLA1	2D	16.1±5.0	18.8±5.5	136.4±6.1
	3D	10.7±3.4	13.6±3.8	128.8±3.6
MLA2	2D	20.5±5.5	23.4±5.9	145.5±6.9
	3D	14.4±4.1	17.6±4.3	137.7±4.0
MLA3	2D	21.2±5.9	25.4±7.3	118.1±7.9
	3D	17.0±4.1	21.8±5.8	117.5±5.5
MLA4	2D	20.3±5.5	23.2±5.9	137.9±7.1
	3D	14.1±3.8	17.1±4.0	128.9±3.9
MLA1b	2D	16.1±4.7	18.9±5.4	123.8±6.2
	3D	11.0±3.6	13.4±4.0	118.4±4.0
MLA2b	2D	19.9±5.1	22.5±5.4	128.6±6.3
	3D	14.6±4.0	17.1±4.1	123.5±4.1
MLA3b	2D	20.6±5.2	24.3±6.3	130.4±7.0
	3D	17.5±4.2	21.8±5.5	129.4±5.1
MLA4b	2D	19.7±5.0	22.4±5.3	121.0±6.6
	3D	13.8±3.7	16.2±3.8	115.1±4.0

Discussion

Despite the increasing interest across research fields in the accurate measurement of foot posture and MLA deformation, the understanding of which skin-marker based model is more suitable to track MLA deformation is still limited. Accuracy of skin-marker based foot kinematics has been seldom reported (Nester et al., 2007), with studies mostly focusing on the source of errors - such as the skin motion artifacts (Schallig et al., 2021; Shultz et al., 2011). The purpose of this study was to assess the accuracy of several skin-marker based models in tracking MLA deformation.

According to the outcome of this study, the measurement of MLA deformation is significantly affected by the skin-marker model; errors between 2 - 10 deg, or 10 - 70% of the nominal radiographic measurements, were observed across all models. While no significant differences were found in accuracy between 2D and 3D definitions, the former showed smaller errors. This can probably be accounted for by the intrinsic planar nature of the radiographic measures used as reference. Models MLA2b and MLA4b showed the lowest errors in measuring MLA deformation across all samples. In these two models, the apex of the MLA angle is the most medial apex of the navicular tuberosity, whereas the sustentaculum tali is used as apex for MLA1 and MLA1b. In fact, Wilcoxon signed-rank tests showed that all b-variations presented smaller errors than the corresponding MLA measures. This seems to indicate that: 1) the marker on the navicular tuberosity is more suitable than the one on the sustentaculum tali to track MLA deformation, and 2) projecting the metatarsal bone marker helps to increase the accuracy in tracking the metatarsal head and thus measure MLA deformation. As far as the latter, visual assessment of the radiographies clearly shows skin-motion artifacts affecting the vertical position of this marker when applying the dorsiflexing wedge. Model MLA1 (2D) showed one of the largest errors in measuring MLA deformation (10 ± 3 deg) and, although not statistically significant, presented a negative correlation with respect to the corresponding radiographic measurements (Table 1). Conversely, MLA2 and MLA2b, and MLA4 and MLA4b showed moderate positive correlations ($R2 = 0.30$-0.42; $p <$

0.05). With respect to the walking trials, the ROM of MLA2b and MLA4b in the stance phase (Table 2) are consistent with what was previously reported using similar definitions (Prachgosin et al., 2015; Tome et al., 2006). Moreover, the present MLA angles in gait (Table 2) are comparable with the data reported by Balsdon et al. (2016) – although on only five subjects - using biplane fluoroscopy for the normal-arched feet group (mean angle = 133.2 ± 7.7 deg).

These results should be interpreted considering some limitations. MLA deformation was estimated in two static postures which were chosen to maximize MLA deformation in controlled static conditions, and to be easily replicated within the X-ray apparatus and in the gait lab. However, while each subject was instructed to maintain a monopodalic full weightbearing posture during the two X- ray acquisitions, the reduced space within the CBCT bore may have slightly affected the total loading applied to the foot with respect to the equivalent unconstrained gait-lab acquisitions. Biplanar video- fluoroscopic analysis associated to 3D reconstruction of the bones relevant to MLA mechanics would be necessary to obtain more accurate dynamic data of medial arch deformation (see e.g. Balsdon et al. 2016). Moreover, the present MLA definitions were limited by the marker-set of the RFM (Leardini et al., 2007; Portinaro et al., 2014); original ad-hoc marker-sets shall be tested in future endeavors.

Although based on static postures, large differences were observed in measuring MLA deformation using different skin-marker based models. While MLA1 proved to be one of the most repeatable across examiners and sessions (Caravaggi et al., 2019), skin-marker models using the marker on the navicular bone as apex of the

MLA angle, and the projection of the metatarsal head marker, appear to be highly preferable to improve accuracy in measuring MLA deformation.

Acknowledgements

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CAPÍTULO IV - A BIOMECÂNICA DO PÉ DURANTE A CORRIDA DE ANTEPÉ E RETROPÉ

Neste capítulo serão apresentados dois artigos originais que avaliaram como o tipo de aterrissagem do pé na corrida (antepé ou retropé) influencia a cinemática dos segmentos pé e as forças e taxas da força na corrida. No primeiro estudo, mostramos que a forma como o pé entra em contato com o solo durante a corrida determina diretamente como será o comportamento cinemático do restante dos segmentos do pé na fase de apoio. O segundo estudo investigou as razões biomecânicas que explicam por que as forças verticais e as taxas de carga de alguns corredores de antepé são similares às de um corredor de retropé.

4.1 Rearfoot, midfoot, and forefoot motion in naturally forefoot and rearfoot strike runners during treadmill running

applied sciences

MDPI

Article

Rearfoot, Midfoot, and Forefoot Motion in Naturally Forefoot and Rearfoot Strike Runners during Treadmill Running

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Abstract

Different location and incidence of lower extremity injuries have been reported in rearfoot strike (RFS) and forefoot strike (FFS) recreational runners. These might be related to functional differences between the two footstrike patterns affecting foot kinematics and thus the incidence of running injuries. The aim of this study was to investigate and compare the kinematic patterns of foot joints between naturally RFS and FFS runners. A validated multi-segment foot model was used to measure 24 foot kinematic variables in long-distance recreational runners while running on a treadmill. These variables included the three-dimensional relative motion between rearfoot, midfoot, and forefoot segments. The footstrike pattern was identified using kinematic data and slow-motion videos. Functional analysis of variance was used to compare the time series of these variables between RFS (n = 49) and FFS (n = 25) runners. In FFS runners, the metatarsal bones were less tilted with respect to the ground, and the metatarsus was less adducted with respect to the calcaneus during stance. In early stance, the calcaneus was more dorsiflexed with respect to the shank and returned to a more plantarflexed position at push-off. FFS runners showed a more adducted calcaneus with respect to the shank and a less inverted midfoot to the calcaneus. The present study has showed that the footstrike angle characterizes foot kinematics in running. These data may help shed more light on the relationship between foot function and running-related injuries.

Keywords: striking pattern; rearfoot strike; forefoot strike; running; multi-segment foot kinematics.

Introduction

The footstrike pattern of runners has received much attention in the past decade, particularly the differences between rearfoot strike (RFS) and midfoot or forefoot strike (FFS). This increasing interest in footstrike patterns can be justified by the possible associations with running-related injuries [1–4]. In particular, FFS has been found to be associated with the attenuation of the lower limbs loads, which helps to reduce the incidence of running-related injuries [4,5]. However, RFS mitigates part of the Achilles tendon loading rate the foot is accountable for, especially in early stance [3].

The effect of striking patterns on lower limb joint kinetics and kinematics has thus far been reported, while foot joint and segment kinematics have rarely been investigated and in a limited number of foot joints and segments only [6–9]. Lower limb kinematics showed that FFS presents the foot angle at initial contact in a plantar flexed position and RFS in dorsiflexed position [10]. In addition, FFS runners contact the ground with greater knee flexion compared to RFS runners [10]. Kinematic differences have also been reported in the frontal plane, where greater rearfoot eversion is observed in FFS compared to RFS [10].

Two recent studies used multi-segment foot models to investigate and compare kinematic patterns in RFS or FFS running. Kelly et al. [7] analyzed the influence of the foot strike technique on medial longitudinal arch mechanics and intrinsic foot muscle function during running. As expected, it was observed that midfoot and rearfoot joint angles of FFS runners were more plantarflexed at footstrike, the rearfoot was less dorsiflexed at mid-stance, and there was more ankle plantarflexion at toe-off compared to the RFS runners. In addition, FFS runners presented a larger loading on the midfoot, which was associated with greater intrinsic foot muscles activation, probably to increase elastic energy storage and return while preventing excessive midfoot deformation. In this study, however, only 13 runners were assessed, and no objective measure was used to determine the runners striking patterns. In fact, runners were instructed to run using an FFS pattern, even if this was not their typical striking pattern. The other study, conducted by Bruening et al. [9], showed that during loading response, the ankle is more plantarflexed, inverted, and adducted in FFS compared to RFS. The midtarsal joint is less inverted and less adducted at initial contact in FFS. During the early stance, the midtarsal has a greater dorsiflexion range and reduced abduction excursion. During the loading phase, the midtarsal is more inverted in FFS and more everted in RFS. The midtarsal joint increases the plantarflexion excursion at late stance in FFS. Finally, the metatarsophalangeal joints are less plantarflexed at early stance, and the dorsiflexion range increases during late stance. However, in this study, the participants were not habitually FFS or RFS, and the authors did not control for potential adaptations that may occur due to the conversion of the footstrike pattern during the data collection session.

We propose another way of analyzing the foot kinematic without immediately resorting to reductionism of variables, analyzing the whole time series using a functional data analysis, instead of a collection of discrete variables within a time series. In this study, we focus on improving our understanding on the effects of the striking pattern on foot kinematics by using a validated multi-segment foot model applied on a large population of naturally RFS and FFS runners. The present results may contribute to further expand our comprehension of the relationships between foot function and running-related injuries, as there are differences in type of injuries reported in FFS and RFS. For example, posterior lower leg injuries are reported for FFS and repetitive stress injuries are reported in RFS [3–5]. Moreover, differences have also been reported in running economy strategies, with FFS being more efficient, therefore resulting in an increased performance compared to RFS during acute transitioning from one pattern to another [11].

In summary, the main limitations of the previous studies were (i) the populations analyzed, these being not naturally FFS and RFS runners; (ii) the lack of control the motor adaptations which occur as runners forcibly convert their footstrike pattern; and (iii) the small sample, the largest size being 18 runners only [9]. Thus, the aim of the present study was to overcome these limitations, investigating and comparing 3D kinematic patterns of foot joints and segments between large populations of naturally RFS and FFS recreational runners. Our hypotheses were: (i) FFS runners land with the more plantarflexed forefoot compared to RFS runners; (ii) first and second metatarsal bones of FFS runners are less inclined with respect to the ground at foot contact due to their landing strategy; and (iii) metatarsal bones of RFS runners present an upward orientation with respect to the ground at initial contact.

Methods

Participants

From a larger ongoing randomized controlled trial, 83 healthy distance runners (age 41.0 ± 6.5 years; running 21.3 ± 14.7 km/week) were assessed. Participants provided informed written consent, and all procedures undertaken were approved by the Ethics Committee of the School of Medicine of the University of São Paulo (protocol number: 031/15). The protocol was previously registered with ClinicalTrials.gov (identifier: NCT02306148). Eligibility criteria included recreational runners between 18 and 55 years old who had been running 20–100 km/week for at least 1 year, with no history of running-related injuries in the 2 months prior to the functional assessment, no experience with minimalist shoes, neutral (normal) feet as determined by the Foot Posture Index, and without chronic diseases or impairments that could influence running performance (e.g., osteoarthritis). An a priori sample size calculation was performed using G*Power [10] considering the averages and standard deviations during late stance of the two groups (RFS and FFS) for the midtarsal flexion/plantarflexion range of motion based on a previous study [9], resulting in a sample size of 72 for 80% power. Due to a larger number of RFS runners compared to FFS, a proportion of 2:1 (RFS:FFS) was used for the sample size calculation [2].

Protocol and Instrumentation

The shank and foot of each participant were outfitted with 16 reflective skin markers (9 mm in diameter) according to the Rizzoli Foot Model (Figure 12) [12,13]. An eight-camera motion analysis system (Vicon Motion System Ltd., Oxford Metrics, Oxford, UK) collected three-dimensional (3D) upright standing and kinematic data during barefoot running at 200 Hz. The participants ran on a force-sensing tandem treadmill at their self-selected speed (AMTI, Watertown, MA, USA) in order to minimize the variability of foot segment motion [14], and were not instructed to use any particular footstrike pattern. A 2–3 min warm-up and familiarization with running barefoot on the treadmill was given to each participant. At least 10 consecutive successful trials with the runners' natural pattern were used in the data analysis.

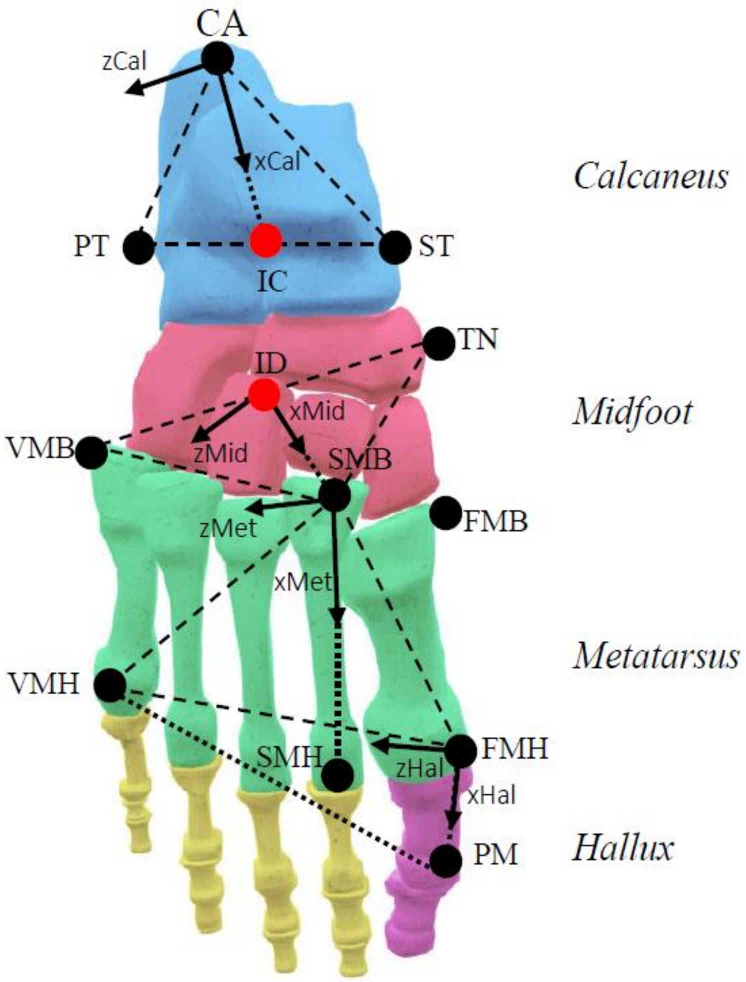

Figure 12 - Diagram of the analyzed foot segments with relevant anatomical landmarks and coordinate reference frames (see Leardini et al. 2007). These were assumed to be rigid segments. CA: upper central ridge of the calcaneus posterior surface, ST: medial apex of the tuberosity of the navicular, VMB: base of the fifth metatarsal, dorsolateral aspect of the fifth

metatarso-cuboid joint, FMB: base of the first metatarsal, dorsomedial aspect of the first metatarso-cuneiform joint, VMH: Head of the fifth metatarsal, dorsolateral aspect of the fifth metatarsophalangeal joint, FMH: head of the first metatarsal, dorsolateral aspect of the first metatarsophalangeal joint, SMH: Head of the second metatarsal, dorsomedial aspect of the second metatarsophalangeal joint, SMB: second metatarsal base, assumed to coincide with the most distal and dorsal aspect of the middle cuneiform, PM: Most distal and dorsal point of the head of the proximal phalanx of the hallux, IC: midpoint of the straight line between PT and ST, ID: midpoint of the straight line between TN and VMB.

Runners were first asked what they believed to be their typical footstrike pattern. This self-reported pattern was verified against kinematic data (e.g., the footstrike angle) and slow-motion sagittal-plane videos from high-speed cameras (120 Hz) (Figure 13). Whenever the footstrike pattern differed from the self-reported one, the runner was asked to maintain its self-reported pattern and was given more time to familiarize with treadmill running. Footstrike angle was defined as the sagittal-plane foot angle relative to the ground at foot contact. The foot angle was defined as shown in Figure 14. Positive foot angles were associated with the RFS pattern and negative foot angles with FFS. Each runner was deemed to be RFS or FFS if the pattern was consistent across more than 30 barefoot strikes out of 50. Those runners who presented a mixed footstrike pattern were excluded from the analysis. Those runners who presented more than 70% of the steps with the same foot strike pattern were included, and 10 consecutive steps presenting the same footstrike pattern were used in the analysis.

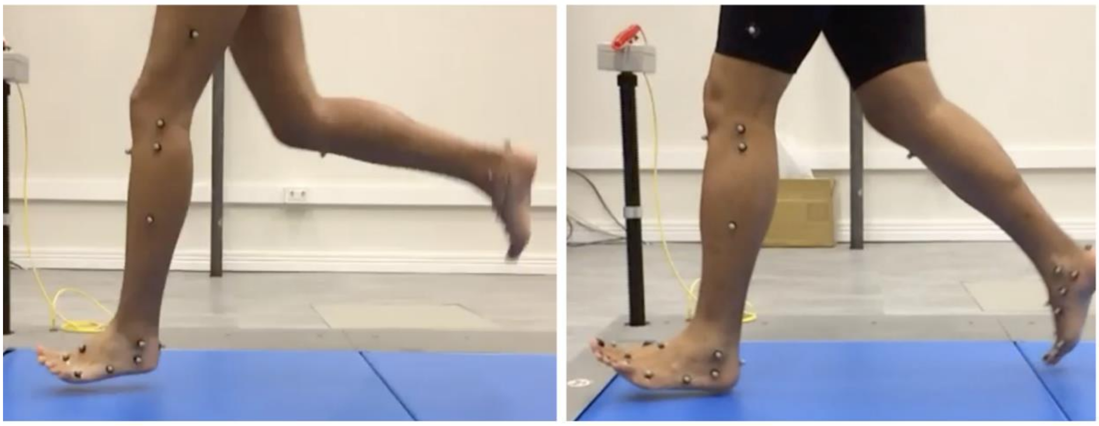

Figure 13 - Illustrations showing the experimental protocol, taken while subjects were running on the force-sensing tandem treadmill at their self-selected speed. Runner's typical footstrike pattern, either forefoot (FFS, left picture) or rearfoot strike (RFS, right picture) pattern, was verified against kinematic data and slow-motion sagittal-plane videos from high-speed cameras.

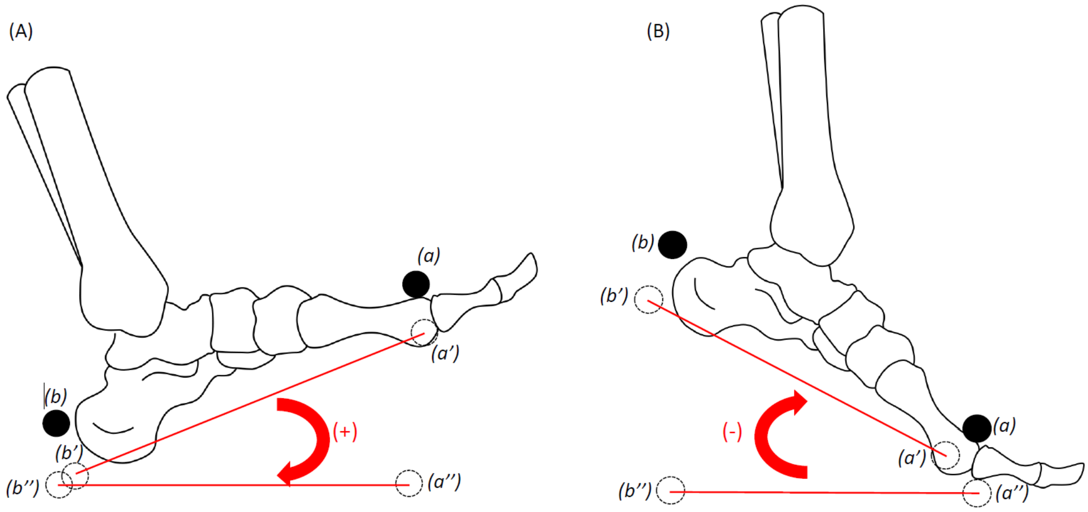

Figure 14 - Foot angle is defined as the angle created by the bisection of a vector from the marker located on the first metatarsal head (a) projected on the foot sole (a') to the marker located on the calcaneus posterior surface (b) projected on the foot sole (b'), with another vector from the marker located on the first metatarsal head (a) projected on the ground (a'') to the marker located on the calcaneus posterior surface (b) projected on the ground (b'') onto the sagittal plane of the foot. Positive foot angles were associated with the (A) rearfoot (RFS) pattern and negative foot angles with (B) forefoot pattern (FFS).

Data Analysis

The Nexus software (version 2.10.3, Vicon, Oxford, UK) was used to reconstruct the 3D coordinates of the skin markers during running. Markers' trajectories were filtered using a Woltring low-pass filter (cutoff frequency = 10 Hz), and processed in Visual3D (C-Motion, Germantown, MD, USA) for joint angles calculation using the joint coordinate system [15]. Accordingly, the present convention for joint rotations was established: dorsi/plantarflexion was assumed to be the rotation about the Z-axis (medio-lateral) of the proximal segment, abduction/adduction the rotation about the Y-axis (vertical) of the distal segment, and eversion/inversion the rotation about the axis orthogonal to the previous two. Ground reaction forces were sampled at 1000 Hz and used to determine stance events and phases. All kinematic data were normalized to the running stance phase duration. The average of each variable across 10 consecutive trials was calculated and used for statistical analysis for each subject. All calculations were performed in Matlab R2015a (MathWorks, Natick, MA, USA).

Rotations about the three axes of the joint coordinate system for the following pairs of segments were analyzed: shank and foot (Sha-Foo); shank and calcaneus (Sha-Cal); calcaneus and midfoot (Cal-Mid); midfoot and metatarsus (Mid-Met), and calcaneus and metatarsus (Cal-Met). Sagittal- and transversal-plane motion between hallux and metatarsus (Met-Hal) were also analyzed. Moreover, sagittal-plane inclination of the first (F2G) second (S2G) and fifth metatarsal bones (V2G) to the ground, and the transverse-plane divergence between first and second metatarsal bones (S2F) and between second and fifth metatarsal bones (S2V) were analyzed. The medial longitudinal arch angle (MLA) was defined as the 3D angle between two vectors, with apex at the marker on the talo-navicular joint and bounded by the markers on the calcaneus (projected on the ground) and on the head of the first metatarsal bone [16].

Groups were compared using the t-test and Chi-squared test or Fisher's exact test for anthropometric, demographic, and training variables (age, Arch Index, and Foot Posture Index) using an alpha level of 0.05. Outcome variables were compared between groups using functional data analysis (FDA), which allow analysis of the full time series of each variable represented by mathematical functions, allowing the analysis of more than a few sets of points. In the functional analysis of variance (fdANOVA), the parameters of the dependent variable are functions, and the design matrix remains a general linear model [17]. This was performed by applying spline bases to the time series before performing fdANOVA using RStudio Software Version 1.2.1335 and the package fdANOVA [18]. For further analysis, we performed fdANOVA to gain a better understanding of the whole movement patterns. Averaged waveforms were time normalized to stance phase, and then means and standard error (SE) bands for each strike pattern were plotted. Mean differences between strike patterns were plotted with 95% confidence interval (CI) bands, which were considered significantly different if $p < 0.05$ and the CI bands did not cross 0.

Results

Nine runners presented a non-identifiable footstrike pattern (mix of RFS and FFS patterns) and were thus removed from the analysis. Of the remaining 74 runners, 49 presented the RFS pattern and 25 the FFS pattern. No differences were found for body mass, height, age, self-selected running speed, or Foot Posture Index between the two groups (Table 5).

Table 5 - Mean (standard deviation) demographics for naturally FFS and RFS runners.

Variable	**RFS (n = 49)**	**FFS (n = 25)**	**Main Effect (p)**
Male/female	23/26	12/13	*p* = 0.563†
Age (year)	41.3 (6.8)	40.4 (6.0)	*p* = 0.558*
Height (m)	1.68 (8.8)	1.70 (9.6)	*p* = 0.198*
Body mass (kg)	68.7 (12.0)	71.7 (14.6)	*p* = 0.348*
Running volume (km/week)	18.9 (14.1)	19.2 (17.8)	*p* = 0.842*
Foot Posture Index (median, min: max)	1, -7: +9	3, -4: +8	*p* = 0.371#
Self-selected speed (km/h)	9.8 (1.2)	9.2 (1.3)	*p* = 0.080*

†Chi-squared test; *t-test; #Fisher's exact test. FFS: forefoot strike, RFS: rearfoot strike. Standard deviations in parentheses.

In the FFS group, a negative foot angle was associated with a forefoot landing strategy, while in the RFS group, positive values were associated with a heel-first landing strategy ($p < 0.001$) (Figure 15). The FFS and RFS groups had a similar dorsiflexion of Sha-Foo at footstrike. At around 15% of stance, the Sha-Foo of FFS runners was slightly more dorsiflexed compared to that of RFS. At around 45% of stance, FFS runners reduced the dorsiflexion and returned to a position of greater plantarflexion compared to RFS runners for the remainder of the stance phase ($p < 0.001$). RFS runners increased Sha-Foo dorsiflexion until 60% of stance, reached a larger dorsiflexion angle, and began plantarflexing later than RFS runners ($p < 0.001$). FFS runners presented a significantly more adducted Sha-Cal ($p = 0.02$) (Figure 15) and a less adducted Cal-Met ($p = 0.01$) with respect to RFS runners during the whole stance phase (Figure 16). The Cal-Mid was significantly less inverted in the FFS group during the whole stance phase ($p = 0.01$) (Figure 16).

F2G and S2G angles presented similar patterns at 20–60% of the stance phase, with RFS runners presenting smaller angles at initial contact. FFS runners presented a significantly larger downward rotation of F2G and S2G after 60% of stance time ($p < 0.001$) (Figure 17). V2G angle was significantly different at initial contact between the two groups ($p < 0.001$). During the first 25% of stance, FFS runners were in a neutral V2G position, while the RFS showed an upward orientation of the V2G. At about 50% of stance, FFS runners returned to a more downward rotation of F2G, S2G, and V2G ($p < 0.01$) (Figure 17). No differences were found in S2F, S2V, MLA, Mid-Met, or Met-Hal angles for all planes between FFS and RFS groups.

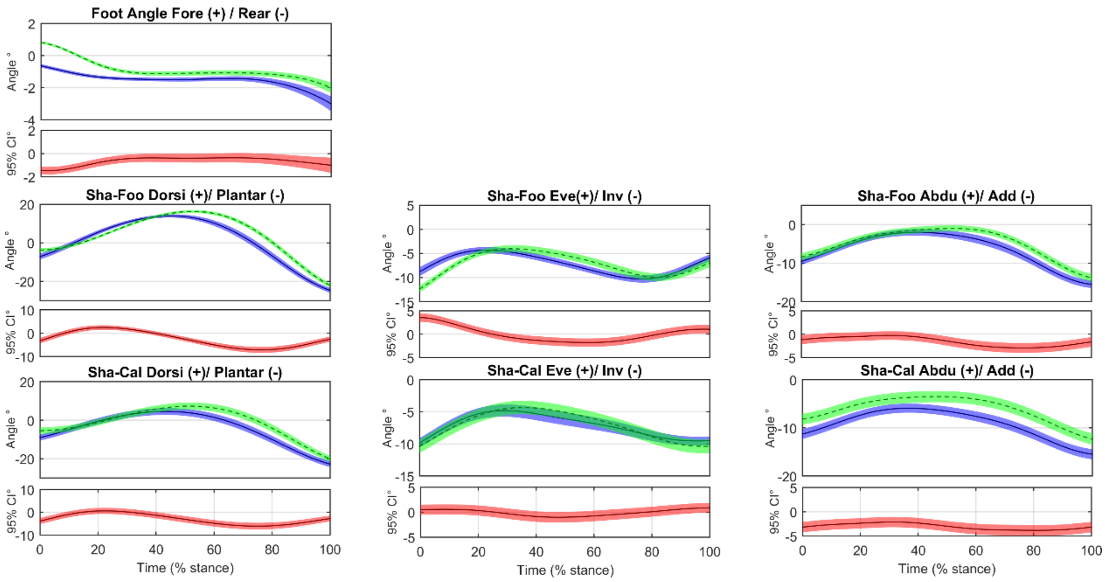

Figure 15 - Foot angle, foot with respect to the shank (Sha-Foo), and calcaneus with respect to the shank (Sha-Cal) joint angles, time normalized across stance phase (0–100%) for forefoot (FFS) and rearfoot (RFS) strike patterns. Each curve contains the mean ± standard error bands (shaded regions). Below each angle plot is a between-condition difference plot (FFS - RFS), containing the mean ± 95% confidence interval bands (95% CI°). Regions where those bands separate from zero can be considered regions of statistical differences. (Legend: FFS in blue, RFS in green)

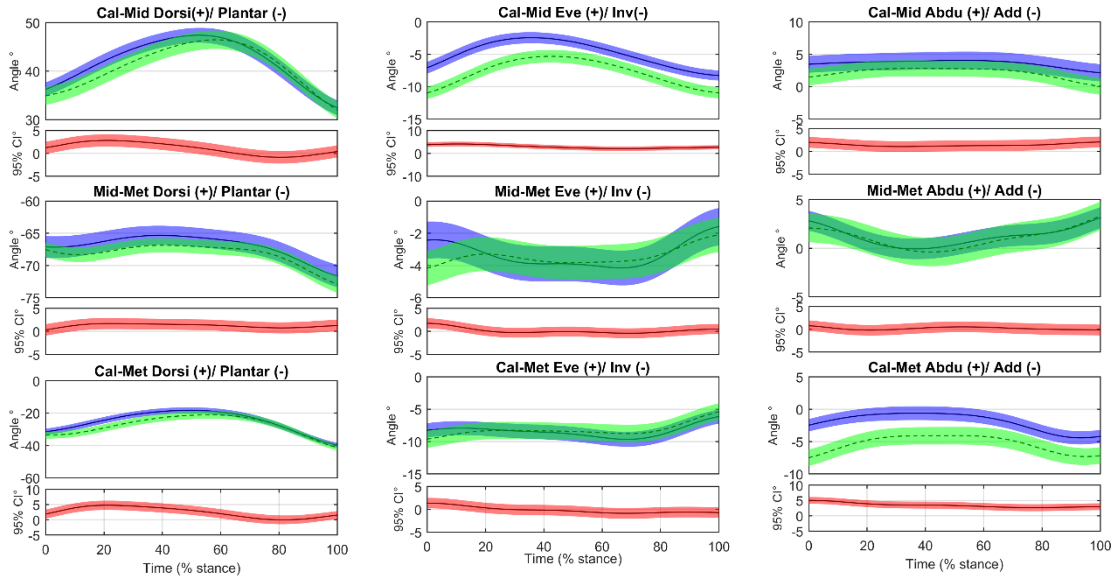

Figure 16 - Midfoot with respect to the calcaneus (Cal-Mid), the metatarsus with respect to the midfoot (Mid-Met), and the metatarsus with respect to the calcaneus (Cal-Met) joint angles, time normalized across stance phase (0–100%) for forefoot (FFS) and rearfoot (RFS) strike patterns. Each curve contains the mean ± standard error bands (shaded regions). Below each angle plot is a between-condition difference plot (FFS - RFS), containing the mean ± 95% confidence interval bands (95% CI°). Regions where those bands separate from zero can be considered regions of statistical differences. (Legend: FFS in blue, RFS in green)

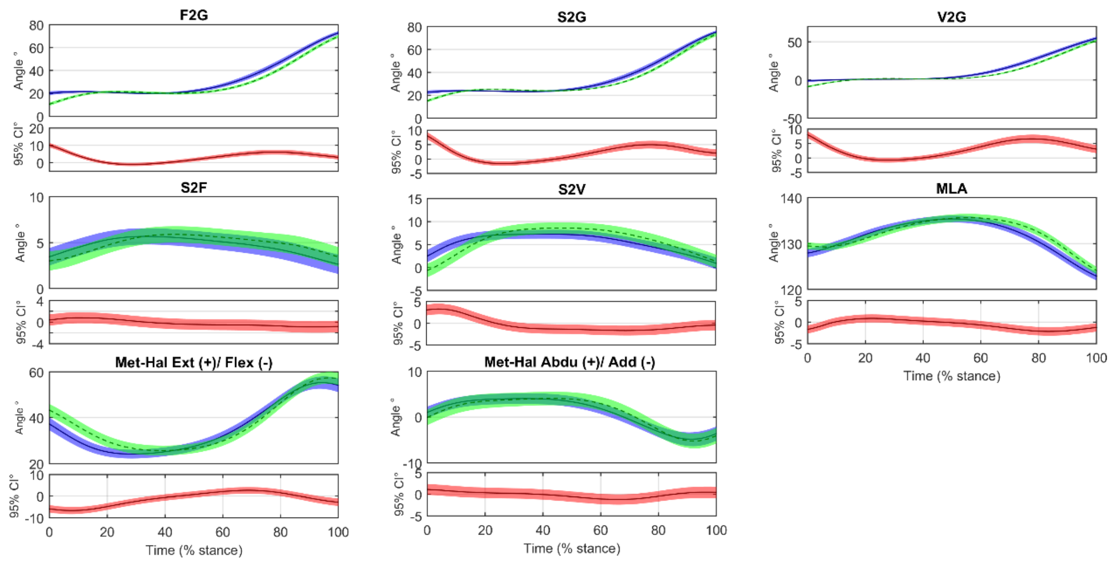

Figure 17 - Sagittal-plane inclination of the 1st metatarsal bone to the ground (F2G), of the 2nd metatarsal bone to the ground (S2G), and also of the 5th metatarsal bone to the ground (V2G); transverse-plane divergence between 1st and 2nd metatarsal bones (S2F); and between 5th and 2nd metatarsal bones (S2V); and first metatarsus and hallux angle (Met-Hal). Time was normalized across stance phase (0–100%) for forefoot (FFS) and rearfoot (RFS) strike patterns. Each curve contains the mean ± standard error bands (shaded regions). Below each

angle plot is a between-condition difference plot (FFS - RFS), containing the mean ± 95% confidence interval bands (95% CI°). Regions where those bands separate from zero can be considered regions of statistical differences. (Legend: FFS in blue, RFS in green)

Discussion

The purpose of this study was to investigate and compare foot joint and segment kinematics between naturally RFS and FFS runners using an established and validated multi-segment kinematic model. In accordance with the first hypothesis, FFS runners landed with the forefoot first and showed an anticipated change in the Sha-Foo plantar/dorsiflexion in the stance phase compared to RFS runners. However, at mid-stance (15–40%), the Sha-Foo of FFS runners was more dorsiflexed, probably due to the larger external ankle dorsiflexion moment associated with this landing strategy [9]. The second hypothesis was also confirmed, as the first and second metatarsal bones in FFS runners was less inclined in relation to the ground (more dorsiflexed) at initial contact. With respect to the third hypothesis, RFS runners showed the fifth metatarsal bone in an upward orientation with respect to the ground and this orientation was combined with a greater adduction of the metatarsus relative to the calcaneus (Cal-Met) at initial contact. These observations are consistent with the theoretical model of an oblique midtarsal joint where plantarflexion is associated with adduction [19–21] when running with an RFS pattern, which was not found in FFS runners.

The larger ankle (i.e., Sha-Foo) plantarflexion at heel strike in FFS runners req uires a higher eccentric activity of the calf muscles. This may lead to Achilles tendinopathy and muscle strain if these soft tissues are not adequately trained for these conditions [10,22]. However, RFS runners are generally more susceptible to injuries due to the higher loads transmitted to the lower limb at heel strike [23,24].

In agreement with Bruening et al. [9], the midfoot of FFS runners was less inverted with respect to the calcaneus (i.e., Cal-Mid joint) throughout the stance phase. In RFS runners, the larger inversion may play an important role in the so-called twisted osteoligamentous plate in the foot structure [25], resulting in an increased resistance to overall foot pronation during the loading phase of running, when this plate tends to untwist, thus providing the necessary protection for the tibiotalar joint from high-impact forces during the stance phase [6,26]. This larger inversion in RFS may also contribute to the conversion of the foot into a sort of rigid lever, which is necessary for effective propulsion during running [27,28].

The calcaneus was more adducted with respect to the shank (i.e., Sha-Cal joint) in the FFS group during the whole stance phase. Fisher et al. [29] showed that calcaneus adduction may be beneficial to the ankle joint coupling, acting as a mechanism to control excessive tibial rotation. This was reported to be associated with patellofemoral pain and iliotibial band syndrome [30], as high relative rotation between the tibia and femur may alter the patella tracking on the distal femur [31]]. The present results seem to suggest a biomechanical mechanism that might explain how a transition from RFS to FFS patterns in gait retraining could reduce patellofemoral pain in runners [32,33] since FFS runners run with a more adducted calcaneus, thus improving ankle joint coupling and tibial rotation control.

The main strengths of this study are the adoption of a reliable and validated skin marker-based multi-segment foot model, which included the tracking of the midfoot and analysis of a large sample of naturally FFS or RFS runners. In particular, by not imposing any footstrike pattern on the participants, we avoided any potential adaptation that might have occurred in converting their footstrike pattern. However, some limitations should be considered when assessing the results of this study. Foot joint kinematics were collected barefoot, which differ from the shod condition; this is indeed difficult to implement using any skin-marker-based multi-segment foot model. However, both groups were subjected to the same testing conditions; thus, if there was any significant change in foot kinematics due to the barefoot condition, this likely affected both groups. Although runners were assessed on a treadmill, the literature shows that most of the spatiotemporal, kinematic, and kinetic parameters are similar between treadmill and overground running [34]. Sinclair et al. [35] examined differences in multi-segment foot kinematics during treadmill and overground running and found only one difference at the ankle and not in the other foot joints: greater plantarflexion at footstrike in treadmill running. Because of the barefoot condition, it is possible that the typical footstrike technique declared by the participant and checked on the treadmill while running barefoot would not be reproduced when running shod in real-world environments. However, Bade et al. [36] suggested that the ability of runners to accurately self-report the footstrike pattern is poor, even when shod.

Despite these limitations, runners could clearly be classified by their strike patterns regarding their foot joint motion. Future prospective studies should address the correlations between running-related injuries and footstrike patterns to potentially improve injury prevention and rehabilitation. The kinematic findings of the present study may be of interest to clinicians and other health professionals to support strategies to prevent or to rehab specific types of injury associated with FFS or RFS, such as plantar fasciitis, Achilles tendinopathy, and metatarsals stress fractures [1], through biomechanics guidance regarding the footstrike patterns.

Conclusions

FFS and RFS runners were found to be characterized by distinct foot joint rotations and bone orientations. In particular, FFS runners landed with the metatarsal bones less tilted with respect to the ground, maintained the metatarsus less adducted during stance and, after landing, maintained the ankle more in dorsiflexion, to return to plantarflexion at push-off. Additionally, FFS runners showed a more adducted calcaneus to the shank and a less inverted midfoot with respect to the calcaneus in the entire stance phase. The present results also brought more evidence on the effect of the footstrike angle on the kinematic pattern of the foot-ankle complex which can be responsible for the differences in running-related injuries, load responses, and running economy strategies between FFS and RFS as largely reported in the literature.

Author Contributions

All authors made substantial contributions to all three of sections. All authors contributed to (1) the conception and design of the study, acquisition of data,

and analysis and interpretation of data; (2) the drafting of the article and critical revision for important intellectual content; and (3) the final approval of the version to be submitted. In the study, these authors took primary responsibility for the following roles: I.C.N.S. was responsible for the study design, interpretation of the data, writing the report, submission of the manuscript, and management. A.B.M. and U.T.T. were responsible for the study design, data collection, analysis, interpretation, writing the report, and submission of the manuscript. P.C. and A.L. were responsible for the study design, interpretation of the data, writing the report, and submission of the manuscript. All authors contributed to the initial draft, revised the manuscript, provided feedback, and approved the final manuscript. All authors have read and agreed to the published version of the manuscript.

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Conflicts of Interest

The authors a rm that this study did not receive any funding/assistance from a commercial organization that could lead to a conflict of interest.

Data Availability

All data access and storage are in keeping with National Health and Medical Research Council guidelines, as approved. All non-confidential files are available from the database published at figshare.com (https://doi.org/10.6084/m9.figshare.10003406.v2). Supported data are available upon request.

Ethics Approval and Consent to Participate

This study was approved by the Ethics Committee of the School of Medicine of the University of São Paulo (18/03/2015, Protocol # 031/15), according to the Declaration of Helsinki Ethical Principles for Medical Research Involving Human Subjects. It was registered at ClinicalTrials.gov (a service of U.S. National Institutes of Health) Identifier NCT02306148 (28 November 2014) under the name “E ects of Foot Strengthening on the Prevalence of Injuries in Long-Distance Runners”. The main researcher explained to each eligible participant every step of the assessment and follow-up, possible risks, and that no compensation or benefits were to be expected. When agreeing to participate, participants were asked for written informed consent, according to standard forms.

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4.2 Not all forefoot strikers are equal

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Abstract

BACKGROUND: Rearfoot strike (RFS) runners typically exhibit an impact peak in their vertical ground reaction force caused by heel impact. This impact is associated with high load rates that have been linked to running injuries. Most of the forefoot strike (FFS) runners do not exhibit this impact peak and have significantly lower load rates compared with RFS runners. However, some FFS runners do exhibit an impact peak and load rates similar to RFS which may be explained by the heel drop after initial contact in FFS. **PURPOSE:** To investigate the relationship between vertical heel kinematics and vertical loads rates in habituated FFS runners. **METHODS:** 49 habitual FFS runners from an ongoing study were included (10F, 39M; age: 35.6±9.3). Ground reaction forces and heel kinematics were collected while the participant ran on an instrumented treadmill at 2.6±0.4m/s. Pearson correlations between average load rate and heel height at initial contact, time to heel contact, heel drop acceleration and heel drop excursion were assessed. These variables were also compared between runners who were chosen based upon exhibiting an impact peak (n=17) and those with the more typical pattern that did not (n=32). **RESULTS:** FFS runners who display vertical impact peaks and high load rates presented with a lower heel height at initial contact, smaller heel drop excursion, shorter time to heel contact and greater heel drop accelerations than those FFS without impacts. Average load rate was significantly (all $p<0.01$) correlated with heel height at initial contact ($r=-0.39$), time to heel contact ($r=-0.68$) and heel drop acceleration ($r=0.58$). A correlation between average loadrate with heel drop excursion ($r=-0.30$, $p=0.038$) was also found. **CONCLUSION:** In habitual FFS runners that presented with vertical impact peaks, we observed a significant relationship between higher vertical load rates (average and instantaneous) and vertical heel kinematics. More specifically, a lower heel at initial contact and smaller heel drop excursion was observed in FFS with high vertical load rates. This kinematic pattern was distinct from FFS without impact peaks. This data suggests that impact peaks in FFS runners may be the result of a shorter window of time in which the plantarflexors can act eccentrically to slow down the heel before impact.

Background

While it is well-recognized that the etiology of running-related injuries is multifactorial, high vertical impact loading appears to be one of the causes (Buist et al., 2007; van der Worp et al., 2015; Davis et al., 2017; Dudley et al., 2017,). It is suggested that being a forefoot strikers (FFS) or midfoot striker decreases the impact peak and the load rate, and, thus, the chance of a running-related injury (Crowell and Davis, 2011; Kulmala et al., 2013; Breen et al., 2015). Hence, the differences between a runner's footstrike pattern, particularly between rearfoot (RFS) and FFS, has received much attention in the past decade (Daoud et al., 2012; Perl et al., 2012; Almeida et al., 2015; Kelly et al., 2018).

Specifically, high vertical load rates (VLRs) have been associated with several common running-related injuries (Milner et al., 2006; Pohl et al., 2009; Davis, 2014; Futrell et al., 2018; Johnson et al., 2020; Tenforde et al., 2020; Johnson and Davis, 2021). The VLR is defined as the slope of the rising force to the impact peak (Crowell and Davis, 2011). In general, higher VLRs are linked to an abrupt and visible vertical impact transient in the ground reaction force caused by harder landings (Samaan et al., 2014).Most habitual FFS runners do not exhibit impact peaks, usually resulting in significantly lower VLRs (Cheung and Davis, 2011; Boyer et al., 2014). A typical ground reaction force curve from a RFS pattern has the characteristic impact peak (Oakley and Pratt, 1988; Lieberman et al., 2010). However, some FFS runners do exhibit impact peaks with associated high VLRs diverging from the typical vertical force curve of a FFS (Boyer et al., 2014; Valenzuela et al., 2014; Rice et al., 2016).

The main difference in kinematic patterns between FFS and RFS runners is the position of the foot related to the ground at initial contact, with FFS runners having a plantarflexed foot angle, and RFS dorsiflexed (Almeida et al., 2015). Bobbert et al. (1991) showed that the impact peak in the running vertical ground reaction force has its origin in the contribution of support leg segments. Recently, Clark et al., (2014, 2017) explored the contribution of the lower limb to explain patterns in vertical ground reaction force waveforms, across different running speeds, foot strike patterns, and footwear types. They demonstrated that the ankle reaches its lowest vertical position earlier for RFS than for FFS. This results in earlier force development from the lower limb mass compared to the mass of the rest of the body leading to an impact transient. Looking at the contribution of the lower limb to the final waveforms, we hypothesized that this method may also explain the presence of impact peaks and high VLRs in FFS runners.

Therefore, the purpose of the present study was to determine whether vertical heel kinematics play a role in the presence of a transient vertical impact peak in FFS runners. Our primary aims were to examine correlations between VLRs and vertical heel kinematics, as well as mean differences with FFS runners who had a vertical impact peak. We hypothesized that higher VLRs and the presence of impact peaks would be associated with higher heel height at initial contact, maximum heel drop acceleration, and heel velocity at initial contact and lower time to heel contact and the heel drop excursion.

Methods

A total of 49 runners were identified (Table 1) from a larger, ongoing study of healthy runners. Inclusion criteria included: forefoot strike runners, barefoot, true minimal or partial minimal runners, ages 18-60 years, running at least 10 miles per week for the past 6 months, and injury free for the past 3 months.

Retroreflective markers were placed based on a standard marker set (Noehren et al., 2013). The heel marker was placed on the lower central ridge of the calcaneus posterior surface. For testing, participants wore shoes that matched the style of their habitual footwear. Participants first warmed up on an instrumented treadmill (AMTI, Watertown, MA), running at comfortable speed for 3 minutes. Following the warm-up, speed was increased to a self-selected pace, defined as a comfortable training pace. Sixteen seconds of ground reaction force data were acquired at 1,5 kHz. Kinematic data were recorded using an eight-camera Vicon MX motion capture system at 250 Hz.

Foot strike pattern was confirmed by visual inspection of high-speed videos (125 fps) capturing force plate contact in the sagittal plane. Force data were filtered using a fourth-order 50-Hz low-pass Butterworth filter in Visual3D (C- motion, Rockville, MD). Foot contact was defined as occurring when vertical force rose above 10 N. Visual 3D was also used to calculate the heel kinematic variables.

The position-time data for the heel marker were used to determine heel height at initial contact (HIC), time to heel contact (THC), maximum heel drop acceleration (HAD), heel drop excursion (HDE) and heel velocity at initial contact (VIC) (figure 18). Vertical average load rate (VALR) and vertical instantaneous load rate (VILR) were calculated from the ground reaction force curve and normalized to body weight (Figure 19). Runners were divided into two groups based on those that exhibited an impact peak and those that did not. The presence of the impact peak in the vertical ground reaction force was verified by visual inspection (i.e., Figure 19).

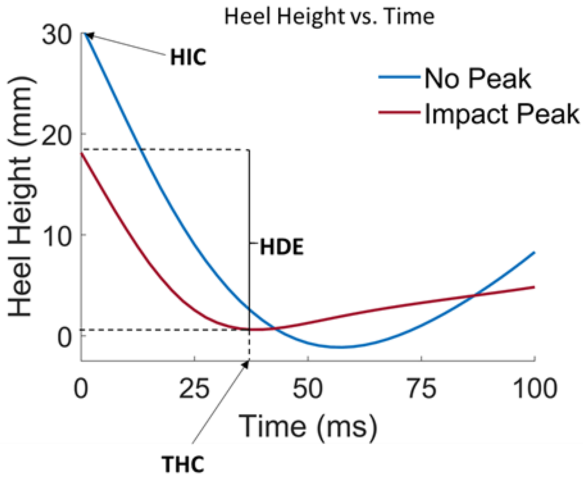

Figure 18 - Graphical representation of Heel height at initial contact (HIC), Time to heel contact (THC), and Heel drop excursion (HDE) according to Clark et al. (2017).

Normality was assessed using Kolmogorov–Smirnov tests. Pearson's correlation coefficients were used to assess the correlation between vertical heel kinematics and VLRs in the group with a vertical impact peak. Correlation strength was assessed using cutoffs proposed by Cohen (2013) ($R \leq 0.2$ = small, $R \leq 0.5$ = moderate, $R > 0.5$ = large). Independent t-tests or Mann-Whitney U tests were used to perform mean comparisons between groups. Effect sizes were calculated and interpreted using Cohen's d (<0.5 = small, 0.5–0.8 = moderate, >0.8 = large)(Cohen, 2013).

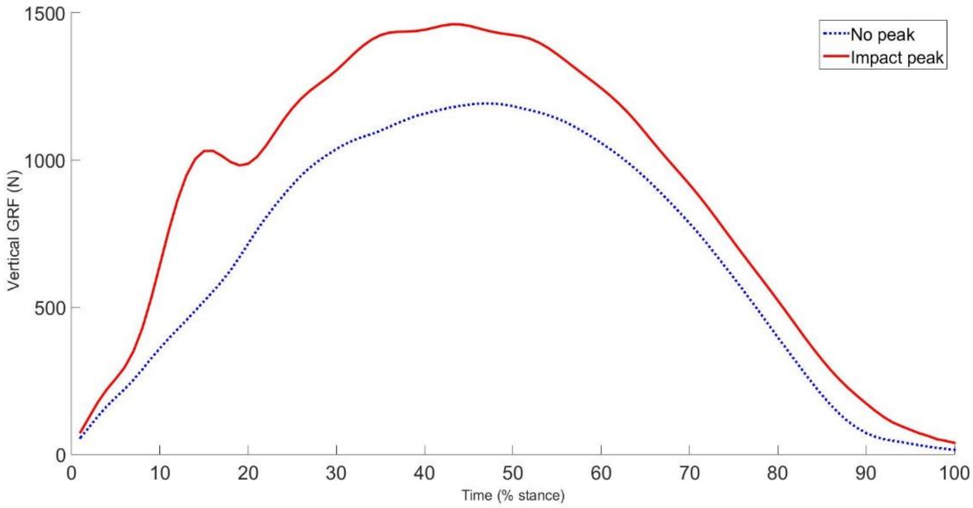

Figure 19 - Ground reaction force waveform from one participant of each group.

Results

An independent samples t-test showed no significant differences in mean in height, body mass or running speed between groups. However, the FFS group who demonstrated an impact peak was slightly older than those without peak (Table 6). Additionally, results of Mann-Whitney U tests demonstrated no significant differences in mean running speed between groups. A sensitivity analysis was run to test for potential confounding effects of the mean difference in age between groups. Age, entered as a covariate in comparisons of vertical heel kinematics between groups, did not significantly alter the main effects.

Table 6 - Mean (SD) demographics for each group.

Variable	Peak	Without Peak	p
Male/female	15/2	24/8	
Age (yr)	42.6 (12.2)	35.4 (7.1)	0.01
Height (m)	1.71 (0.05)	1.78 (0.08)	0.67
Body mass (kg)	71.9 (9.1)	73.1 (9.2)	0.76
Speed (m/s)	2.5 (0.3)	2.7 (0.4)	0.92
Bar/TM/PM	6/13/14	9/1/6	

Abbreviations: Bar, barefoot; TM, True Minimal shoe; PM, Partial minimal shoe

When analyzing the two groups as a whole, we observed the strongest correlation between time to heel contact and VALR (r=0.68; p<0.001) (table 7). A large correlation was noted between maximum heel drop acceleration and VALR (r=0.58; p<0.001) (Table 7). Weaker correlations were found for VALR and heel drop excursion (r=0.30; p=0.038), as well as heel height at initial contact (r=0.39; p=0.006). There was not a significant correlation between heel velocity at initial contact and VALR (r=0.08; p=0.58) (Table 7). Finally, correlations with heel kinematic variables were similar for VALR and VILR (Table 8).

Table 7 - Correlation between the vertical average load rate (VALR), vertical instantaneous load rate (VILR) and heel height at initial contact (HIC), time to heel contact (THC), maximum heel drop acceleration (HAD), heel drop excursion (HDE) and heel velocity at initial contact (VIC). * Statistically significant (P<0.05).

	VALR (BW/s)		**VILR (BW/s)**	
Variable	*r*	*p-value*	*r*	*p-value*
HIC (mm)	-0.39	0.006*	-0.39	0.006*
THC (ms)	-0.68	<0.001*	-0.65	<0.001*
had (m/s/s)	0.58	<0.001*	0.59	<0.001*
HDE (mm)	-0.30	0.038*	-0.30	0.065
VIC (m/s)	-0.08	0.58	-0.08	0.55

FFS runners with an impact peak exhibited significantly higher VALR (p<0.001) and VILR (p<0.001) (table 8). Additionally, they demonstrated a lower heel height at initial contact (HIC) (p<0.001), a shorter time to heel contact (THC) (p<0.001) and higher max heel drop acceleration (HAD) (p=0.010). The FFS runners with an impact peak also exhibited significantly lower heel drop excursion (HDE) (p=0.005) compared to those without impact peaks. There was no difference in heel velocity (VIC) during initial contact between FFS with impact peak and those without impact peak (p=0.42).

Table 8 - Group mean (SD) values for vertical load rates and heel kinematic variables

	No Impact Peaks	**Impact Peaks**	**p**
n	33	16	
VALR (BW/s)	35.2 (9.9)	53.8 (15.6)	<0.001*
VILR (BW/s)	47.3 (13.8)	80.3 (22.8)	<0.001*
HIC (mm)	19.5 (12.3)	5.5 (9.0)	<0.001*
THC (ms)	56.4 (11.9)	36.8 (8.9)	<0.001*
HDA (m/s/s)	17.5 (4.7)	21.6 (5.4)	0.010*
HDE (mm)	19.8 (8.7)	12.5 (6.6)	0.005*
VIC (m/s)	-0.6 (0.2)	-0.5 (0.2)	0.42

Abbreviations: VALR: vertical average load rate, VILR: vertical instantaneous load rate, HIC: heel height at initial contact, THC: time to heel contact, HAD: maximum heel drop acceleration, HDE: heel drop excursion, and VIC: heel velocity at initial contact *Significant difference between groups, p< 0.05.

Discussion

This study aimed to determine the relationship between heel kinematics relative to the ground (vertical axis) and vertical load rates in habituated FFS runners. We hypothesized that runners with higher VLR would present with a higher heel height at initial contact, shorter time to heel contact, higher max heel drop acceleration, lower heel drop excursion and heel higher velocity at initial contact. We also hypothesized that significant differences in the heel kinematics and VLR would be found when compared to those FFS runners with a non-visible impact peak.

Our first hypothesis was partially supported. Although we observed a significant correlation between heel height at initial contact, time to heel contact, heel drop excursion and heel drop acceleration with both VLRs, no association was found between heel velocity at initial contact with either VALR or VILR in FFS runners. In addition, our hypothesis that significant differences in the heel kinematics and VLR would be found between FFS runners with and without transient impact peaks was also partially supported. Again, the velocity at initial contact was the only variable that was not significantly different between groups. This suggests that, for both groups, an individual's heel is approaching the ground at the similar speed. However, FFS runners without visible impact peaks are taking a longer period of time to slow down the body's mass (lower acceleration), compared to those with impact peaks. This is supported by the time to heel contact being greater in the FFS without visible impact peaks. Our findings are in agreement with Clark et al. (2017) that found shorter time to heel contact in those runners (RFS) with a visible impact peak.

FFS runners require stronger calf muscles to eccentrically control ankle dorsiflexion at initial contact (Williams and McClay, 2000, Yong et al., 2014). Our results suggest that in FFS runners that do not exhibit impact peaks, these plantarflexors are acting with adequate control and more efficiently in terms of time of activation and power output. In turn, this results in the reduction of vertical forces during early stance, potentially obscuring the vertical impact peak (Hamill and Gruber, 2017). Our results also suggest that a lower heel position at initial contact in

FFS runners that do exhibit impact peaks results in less time for the plantarflexors to act eccentrically to slow down the heel. These runners show a shorter time to heel descent and a higher heel drop acceleration compared to the FFS runners without impact peaks, resulting in greater VLRs. In agreement with our results, Gruber et al. (2017) demonstrated that runners that do not exhibit a prominent (visible) impact peak decelerate the lower limbs later after initial ground contact compared with RFS runners that do. Thus, when a FFS runners heel contacts the ground with a high acceleration, the impact peak is not hidden by the active peak resulting from the acceleration of whole body (Hamill and Gruber, 2017).

While there is some suggestion that running with a FFS pattern may reduce injury, it is important to note that transitioning to a FFS is not without risk. By definition, FFS pattern has the foot more plantarflexed at contact compared to RFS. However, Rice et al. (2016) found that running with exaggerated (higher) plantarflexion may result in high braking and medio-lateral forces in early stance and could also lead to higher VLRs. On the other hand, running with too little plantarflexion may result in not enough time for the calf to decelerate the landing through eccentric action, resulting in impact peaks.

Limitations with the present study include the uneven distribution of participants between the groups resulting in reduced statistical power for some of the comparisons. Additionally, although the same category of running shoe was used in the lab to abate any effect of shoe type, this still may have induced some unnatural changes in their loading (Hunter et al., 2020).

In conclusion, this is the first study to investigate the relationship between vertical heel kinematics and high VLRs in habituated FFS runners with a presence of impact peaks. We observed a significant relationship between higher VLRs (average and instantaneous) with a lower heel at initial contact and smaller heel drop excursion. This kinematic pattern likely contributes to a longer time to heel contact. This suggests that a significantly different heel contact pattern, at least partially explains the presence of impact peaks and high VLRs in FFS runners. Training FFS runners to control their heel after foot strike may be a viable method for lowering VLRs in this group, thereby potentially lowering their risk for a running-related injury

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CAPÍTULO V - EFICÁCIA DE UMA INTERVENÇÃO TERAPÊUTICA PARA OS PÉS NA BIOMECÂNICA DA CORRIDA

Neste capítulo é apresentado o artigo que avaliou a eficácia de uma intervenção terapêutica para os pés, na cinemática do pé e nas forças e taxas de carga de corredores recreacionais de longa distância.

5.1 Effects of a foot-core training on foot-ankle kinematics and running kinetics in runners: secondary outcomes from a randomized controlled trial

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Abstract

This study investigated the effectiveness of an 8-week foot-core exercise training program on foot-ankle kinematics and running kinetics, with particular interest on biomechanical outcomes considered as risk factors for running-related injuries in recreational runners. A single-blind randomized controlled trial was conducted with 87 recreational runners randomly allocated in either control (CG) or intervention group (IG) assessed at baseline and after 8 weeks of foot-core training. The IG underwent the foot-core training 3-times/week while the CG followed a placebo lower limb stretching

protocol. The participants ran on a force-instrumented treadmill at a self-selected speed, while foot segment motion was captured simultaneously with kinetic measurements. After the intervention, compared to CG, IG strike the ground with a more inverted calcaneus and a less dorsiflexed midfoot; at midstance, ran with a less plantarflexed and more adducted forefoot, and a more abducted hallux; and at the push-off, IG ran with a less dorsiflexed midfoot, a less adducted and more dorsiflexed hallux. The IG runners also decreased the medial longitudinal arch excursion and increased the rearfoot inversion. The 8-week foot-core exercise program had no effect on impact and breaking forces or on loading rates, however it was effective to change foot-ankle kinematic patterns.

Introduction

Running is one of the most popular and practiced sports/fitness activities worldwide due to its simple requirements in terms of gear and indoor/outdoor environment. However, one of the drawbacks is the high incidence of running-related injuries (RRI) [1,2]. The etiology of RRI is believed to be multifactorial[3–6], and the following biomechanical risk factors are usually listed: altered medial longitudinal arch (MLA) posture[7–13], greater ankle[14–16] or rearfoot[17–20] eversion, higher loading rates[21–25], impact peaks[21–23,26] and breaking forces[27–29].

Several therapeutic strategies have been implemented in the last decades to minimize RRI incidence, however these have yielded poor outcomes[30–32]. Some of the most commonly adopted therapeutic approaches to reduce RRI are strengthening programs focused on the hip and the core areas – i.e. the “top-down” approach[33–35].

This approach claims that an increased hip and core muscle strength would contribute to the reduction of non-sagittal joint movements and moments, and thus of the loads in the adjacent joints in the lower limbs, which in turn would result in lower risks of RRI[33–38]. Although this approach is very popular, evidence of its beneficial effects in diminishing the incidence and the biomechanical risk factors of RRI[39] over the last few decades are yet to be proven[40].

On the other hand, the so-called “bottom-up” approach can be a promising strategy that focuses on foot-ankle strengthening programs that, according to its theoretical assumptions[40–44], may potentially change the mechanical/biomechanical response of more proximal joints (knee, hip). The foot is a biomechanically complex structure made of 26 bones, four layers of plantar intrinsic muscles and several joints providing the foot with multiple degrees of freedom. Active and passive elements in the foot, such as ligaments and soft tissues, act in synergy to make the foot a mobile adapter capable of receiving and attenuating external loads, and of storing and releasing elastic energy[45,46]. Therefore, the bottom-up approach hypothesis is that a stronger and more functional foot may potentially enhance the body’s ability to tolerate foot impacts, reduce internal loads of foot joints due the small lever arms of the intrinsic muscles, which in turn would decrease RRI[40,42,47–50].

There is some evidence that this approach is effective in preventing RRI and promoting functional gains related to running. A previous proof-of-concept study developed by our group has shown that an 8-week foot-core strengthening program increased the intrinsic foot muscle anatomical cross-sectional area and propulsive impulse during running in recreational long- and middle-distance runners[51]. In addition, our previous randomized trial showed a 2.42-fold reduction of RRI incidence at one year follow-up of a foot-core training in healthy runners compared to a stretching placebo program[52]. In this study, following the successful results of our previous single blind, randomized controlled trial concerning RRI prevention, we would like to further assess the effectiveness of the 8-week foot-core exercise training program[52] on foot-ankle kinematics and running kinetics, with particular interest on biomechanical outcomes considered as risk factors for RRI in recreational runners.

Results and discussion

Our study aimed to evaluate the effectiveness of foot-core exercise training on foot kinematics, running kinetics and on biomechanical risk-factor outcomes for RRI in recreational runners. The results (tables 9 and 10, and figures 20 and 21) and discussion are organized and structured for both discrete and continuous analysis applied to the biomechanical data acquired in this randomized clinical trial (RCT).

Baseline assessment data are described in table 9. The participants were randomly assigned either to the control group or to the 8-week supervised foot core training group (Figure 20). They were on average 40.3 (SD 6.9) years old, and the majority (51.2%) was female, with a mean running experience of 7.1 (SD 6.2) years, Foot Posture Index (FPI) 2.0 as a median (25th percentile = -2.25 and 75th percentile = 4.0, 8% highly supinated, 26% supinated, 49% normal, 14% pronated, 1% highly pronated). 43.5% of the runners reported a RRI in the 12 months prior to the participation in the study (Table 9). Participants were recommended to maintain their running routine during the study period, which was closely monitored to ensure that no significant difference was present between subjects (mean volume of 83.06 (SD 59.58) km/week) and between groups during the 8-week intervention period (IG = 87.78 (SD 60.56) km/week, CG = 87.73 (SD 56.16) km/week).

During the 8-week training program, all participants completed a custom online survey regarding new RRI (if any had occurred) and completed the remote training sessions. The dropout rate, was 4.9% (2 participants) in the IG, and 2.2% (1 participant) in the CG ; the dropout reasons are described elsewhere[52]. The adherence to the IG protocol was measured as the attendance to the locally supervised training by the intervention group. Participants were expected to attend the training with the designated researcher once a week. The full protocol lasted 8 weeks, and participants were excluded if missing two consecutive weekly sessions. The total adherence to the protocol was 96.7 %, where 100% corresponds to all participants attending all sessions (n= 304, after excluding 3 injured participants). Of the 87 runners, 20 sustained a RRI at 1-year follow-up: 6/41 in the IG, and 14/46 in the CG (Figure 22). Injuries in the IG were shin splint, plantar fasciitis and calcaneal tendinitis. Injuries in the CG were patellofemoral pain, shin splint and thigh strain.

Table 9 - Baseline characteristics of participants from the intervention and control groups.

	All participants		Intervention group		Control group	
	N	%/ Mean (SD)	N	%/ Mean (SD)	N	%/ Mean (SD)
N	87		41	47.1%	46	52.9%
Demographics						
Sex (male)	42	48.8%	17	41.5%	25	54.3%
Age (years)		40.3 (6.9)		40.3 (7.7)		40.3 (6.1)
Body mass (kg)		70.5 (13.1)		67.2 (12.1)		73.5 (13.0)
Height (m)		169.3 (8.8)		166.5 (7.6)		171.8 (9.0)
Body Mass Index (kg/m²)		24.5 (3.2)		24.1 (3.0)		24.8 (3.0)
Training						
Running Experience (years)		6.5 (5.7)		5.9 (5.1)		7.1 (6.2)
Running frequency per week		3.7(1.0)		3.8(1.0)		3.6(1.2)
Running volume per week (km)		35.8 (27.6)		31.7 (22.5)		39.4 (30.8)
Average pace (min/km)		6.58” (1.36)		6.46” (2.36)		6.69” (2.38)
Running event						
Member of athletic association (yes)	38	43.7%	19	46.3%	19	41.3%
Participated in a running event before (yes)	83	95.4%	40	97.6%	43	93.5%
Number of running events before		37.0 (41.7)		29.3 (31.8)		44.0 (47.5)
Anthropometrics						
Foot posture index – median (25th and 75th percentiles)		2.0 (-2.25; 4.0)		2.0 (-3.0; 4.0)		1.0 (-1.0; 4.0)
Cavanagh & Rodgers index (right foot)		0.20 (0.06)		0.21 (0.06)		0.21 (0.05)
Previous RRI						
Previous RRI in previous 12 months (yes)	40	46.0%	20	48.8%	20	43.5%
FHSQ score (0-100 points)						
Foot pain		90.5 (12.7)		89.9 (13.3)		91.6 (12.0)
Foot function		98.2 (6.0)		98.8 (5.0)		97.6 (6.6)
Shoes		74.5 (24.8)		73.4 (26.9)		76.8 (22.3)
General Foot Health		78.4 (22.9)		76.4(25.0)		80.3 (20.4)
General Health		86.2 (13.4)		87.1 (12.9)		85.1 (13.6)
Physical Activity		95.5 (15.3)		95.1 (15.0)		95. 8 (15.3)
Social Activity		87.5 (15.0)		88.4 (14.2)		86.7 (15.5)
Vigor		75.2 (13.5)		74.1 (11.8)		76.1 (14.4)
Running Biomechanics						
Medial Longitudinal Arch ROM (deg)		3.40 (7.39)		6.16 (8.14)		3.59 (7.89)
Sha-Cal Inv (-) Peak (deg)		-3.12 (7.38)		-0.56 (7.42)		-3.30 (8.71)
Sha-Cal Eve (+) Peak (deg)		6.81 (2.82)		6.72 (3.29)		6.89 (2.35)
Vertical Impact Peak (BW)		1.14 (0.49)		1.13 (0.39)		1.21 (0.44)
Vertical Average Load Rate (BW*s^{-1})		75.05 (55.75)		75.19 (46.43)		73.48 (43.17)
Peak Braking Force (BW)		-0.24 (0.06)		-0.24 (0.05)		-0.24 (0.05)

Abbreviations: ROM: range of motion, BW: bodyweight, Eve: eversion, Inv: inversion, RRI: running-related injury

The effect of the foot-core training on discrete biomechanical risk factors for RRI

The average loading rate, impact and breaking force peaks were not significantly different between IG and CG at the 8-week follow-up (table 10). Historically, these parameters have been retrospectively and prospectively associated to RRI thus were here chosen as secondary outcomes. Retrospective studies have shown a strong association between higher vertical loading rates and tibial shock with stress fractures in female runners[23], greater vertical impact forces and loading rates with overuse RRI[53] and higher breaking forces in female runners who sustained an injury in a 15-week period[27]. Furthermore, in prospective studies, higher impacts and loading rates were observed in runners who sustained RRI in a 2-year period[21] and in novice male runners who sustained an injury in a 9-week period[25]. However, another prospective study did not find differences in loading rates between injured and uninjured collegiate runners at 12-week follow-up[54]. The etiology of RRI is multifactorial and the different types of RRI observed in our RCT after 1-year period (14 in the CG and 6 in the IG) were probably generated by multiple RRI mechanisms. However, establishing a direct relationship between this kinetic risk factor and RRIs would be difficult due to the small sample and of the low statistical power.

Table 10 - Mean pre- and post-intervention values for kinetic and kinematic biomechanical measures in the experimental groups. P-values of the interaction effect (group × time) are presented. * Indicates significant differences.

Variable	Intervention Group		Control Group		Interaction
	Pre	*Post*	*Pre*	*Post*	*P*
MLA ROM (deg)	6.16 (8.14)	0.16 (6.75)	3.59 (7.89)	2.88 (5.36)	0.024 *
Sha-Cal Inv (-) Peak (deg)	-0.56 (7.42)	-5.74 (6.31)	-3.30 (8.71)	-3.51 (5.70)	0.037 *
Sha-Cal Eve (+) Peak (deg)	6.72 (3.29)	5.90 (2.95)	6.89 (2.35)	6.39 (1.88)	0.557
Vertical Impact Peak (BW)	1.13 (0.39)	1.14 (0.55)	1.21 (0.44)	1.09 (0.58)	0.129
Vertical Average Load Rate ($BW*s^{-1}$)	75.19 (46.43)	77.17 (57.52)	73.48 (43.17)	72.84 (65.73)	0.537
Peak braking Force (BW)	-0.24 (0.05)	-0.24 (0.07)	-0.24 (0.05)	-0.24 (0.05)	0.934

Despite the IG being 2.42 times less likely to experience a RRI within the 12-month study period following the foot-core intervention[52], no reduction in loading rates - which is considered a biomechanical-related risk factor[21,25,27,53] - was observed after 8 weeks(table 10). In the present study, although changes in running biomechanics were assessed at 8 weeks of intervention, the RRI incidence was followed over 12 months, and this could possibly hinder a potential association between any kinetic-related risk factors that was only assessed at 8-week, to the RRI incidence that was assessed at 1-year follow-up. Future studies should further evaluate the effects of specific foot-ankle intervention strategies on the modification of the loading variables associated with RRIs throughout the full trial period and its relationship with the reduction of RRI risk.

A significant reduction in the MLA ROM (p=0.024) was observed after 8-week foot-core training in the IG compared to the CG (table 10). The foot-core training strengthened some of the intrinsic foot muscles[55] responsible for sustaining the MLA[56], possibly increasing the resistance to its deformation during running and thus resulting in a smaller amount of arch collapse in the IG. This is consistent with what reported by

Mulligan and Cook [57] who found a decreased navicular drop after a 4-week intrinsic foot muscle training program. The MLA should have the capacity to be flexible in response to the running loads , allowing foot joint adjustments to dampen impacts through multiple mechanisms, including stiffness and power absorption, but it must also be rigid enough to allow propulsion in the push-off phase[58]. Our foot-core training may have increased the ability of the plantar intrinsic muscles to provide force-dependent alterations in the MLA stiffness and facilitate efficient foot-to-ground contact during running[45,59]. An actively-restricted MLA may help decreasing the mechanical demand on the foot soft tissues, such as ligaments, fascia and tendons and may result in less injuries in these structures[9], such as plantar fasciitis which derives from repetitive abnormal strain and loading of plantar fascia and flattening of the MLA[60,61].

Runners with high MLA had a greater incidence of ankle, bony and lateral sprain injuries, whereas those with lower MLA exhibited more knee, soft tissue and medial sprain injuries[9]. A further mediation analysis could reveal if the changes observed in MLA behavior in the IG is associated to the reduction of RRI in our RCT[52]. Further research should be conducted to determine how the changes in MLA pattern observed after the training program modify the running performance, since our previous proof-of-concept study showed that the foot-core training increased the vertical impulse during running[55].

The 8-week foot-core training affected the rearfoot inversion peak (Sha-Cal angle) (table 10) as the rearfoot presented increased inversion ($p=0.037$) to the shank with respect to what observed in controls (table 10). We may speculate that this is a consequence of the strengthening of the extrinsic foot-ankle muscles, such as the tibialis posterior, thus promoting the inversion of the calcaneus and resisting eversion during stance phase[62–64] and stabilizing the MLA[13,65]. The pathomechanics of medial tibial stress syndrome caused by periosteum inflammation is probably linked to an excessive fascial traction caused by muscle tension resulting from excessive and/or prolonged pronation. A more inverted calcaneus in the IG may have increased the twisting of the osteoligamentous plate at initial ground contact[66], which could consequently increase the resistance to pronation during the loading phase of running, when this plate tends to untwist. This increased resistance to calcaneus pronation in the IG may have provided the necessary protection for the tibiotalar joint from high traction forces imposed by the evertors and invertors muscles during the stance phase[67], resulting in less chance for occurring an injury in the IG compared to the CG, who presented more lower leg RRI[52].

In order to better describe/measure the complexity of the interaction between foot joints in running, we analyzed the changes in 24 kinematic time-series from the Rizzoli Foot Model [68], we next explored changes resulting from the intervention in the 24 kinematic time series from the Rizzoli Foot model. We performed a vector analysis of the resultant angles using 1D-Statistical parametric mapping (1D-SPM) to compare the CG and IG. This approach does not rely on the experimenter subjective selection of the appropriate discrete variables, allowing to identify changes in the whole time series that may have been missed using a discrete-parameter approach.

The effect of the foot-core training on foot-ankle kinematic pattern during whole stance phase of running

Inclination of the metatarsal bones to the ground (F2G, S2G, V2G) and metatarsal bones divergence in the transverse plane of the foot (S2F and S2V) were not different between IG and CG after 8 weeks [see Additional file 2 – Figure 23]. Despite our previous proof-of-concept study[55] showing that the foot-core training increased the muscle volume of the abductor digiti minimi and flexor digitorum brevis, the full RCT did not result in changes in the kinematics of the metatarsal bones. It was expected an increase of F2G S2G and V2G as the MLA is raising and shortening. However, these changes may be very small and difficult to detect with skin-markers.

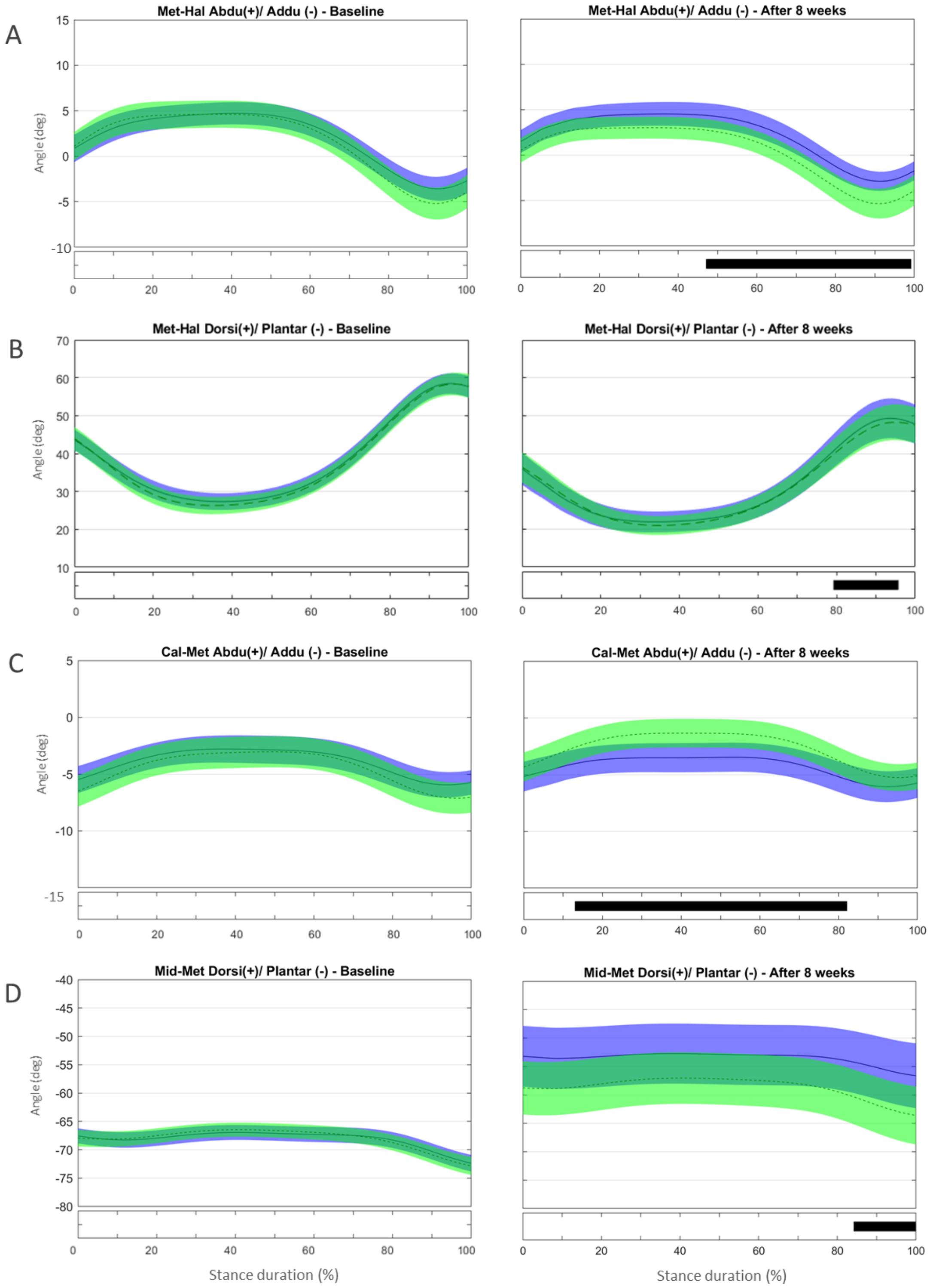

Figure 20 - Mean (±1SD) joint rotation angles during normalized stance phase duration of running in the CG (left) and IG (right). From top to bottom: transverse-plane rotations between hallux and metatarsus (A); sagittal-plane rotations between hallux and metatarsus (B); transverse-plane rotations between metatarsus and calcaneus (C), and sagittal-plane rotations between metatarsus and midfoot (D). Green, CG group; Blue, IG group. The black bar below the graph represents the time during which the differences between the groups occurred ($p<0.05$), what was indicated by the SPM{t} statistic.

No difference was observed in Mid-Met and Cal-Met kinematics between IG and CG at baseline. The intervention had an effect in the sagittal-plane motion of Mid-Met by reducing metatarsal bones plantarflexion from 84 to 100% of stance compared to controls (t*2.764, p=0.016) (Figure 20D). After the foot-core training, the reduction in the Mid-Met plantarflexion toward a dorsiflexion from 84% to push off may be a consequence of a more fixed position of the midfoot and metatarsal bones to the ground at the push-off. The position of the midfoot (Cal-Mid) at the push off may have influenced the metatarsus segment (Mid-Met) as in a closed kinetic chain leading this segment to move in the same direction of the midfoot[72],and thus resulting in a more dorsiflexion of the first metatarsus-phalangeal joint (Met-Hal) as discussed previously.

After the training, the reduction in the Mid-Met plantarflexion toward a dorsiflexion from 84% of stance After the intervention, the IG increased the Cal-Met adduction from 13 to 82% of stance compared to the CG (t*2.722, p=0.008) (Figure 20C). The concomitant reduction in MLA ROM (discrete analysis) seems to show that the IG presents a "stiffer" foot that behaves like a rigid lever, allowing a greater plantarflexion torque to be transmitted to the ground during running[73]. Further research is needed to determine how these changes in the MLA and Cal-Met patterns affect the running performance. Although Messier et al.[18] did not find differences in forefoot adduction between injured and uninjured runners in a 2-year prospective study, the present foot-core training changed the metatarsus position and motion during running; a mediation effect analysis should be performed to assess how this change was related to the lower RRI incidence observed in the IG [52].

At baseline, no difference was observed in frontal- and sagittal-plane midfoot to calcaneus angles (Cal-Mid) between IG and CG. After the training program, the IG presented a reduced Cal-Mid dorsiflexion at early stance (0-20% of stance; t*2.820, p=0.014) and at push-off (80-100% of stance; t*2.820, p=0.013) compared to controls (Figure 21B). The latter presented lower Cal-Mid inversion from 25 to 45% of stance (t*2.704, p=0.014) after 8 weeks [see Additional file 2 – Figure 24].

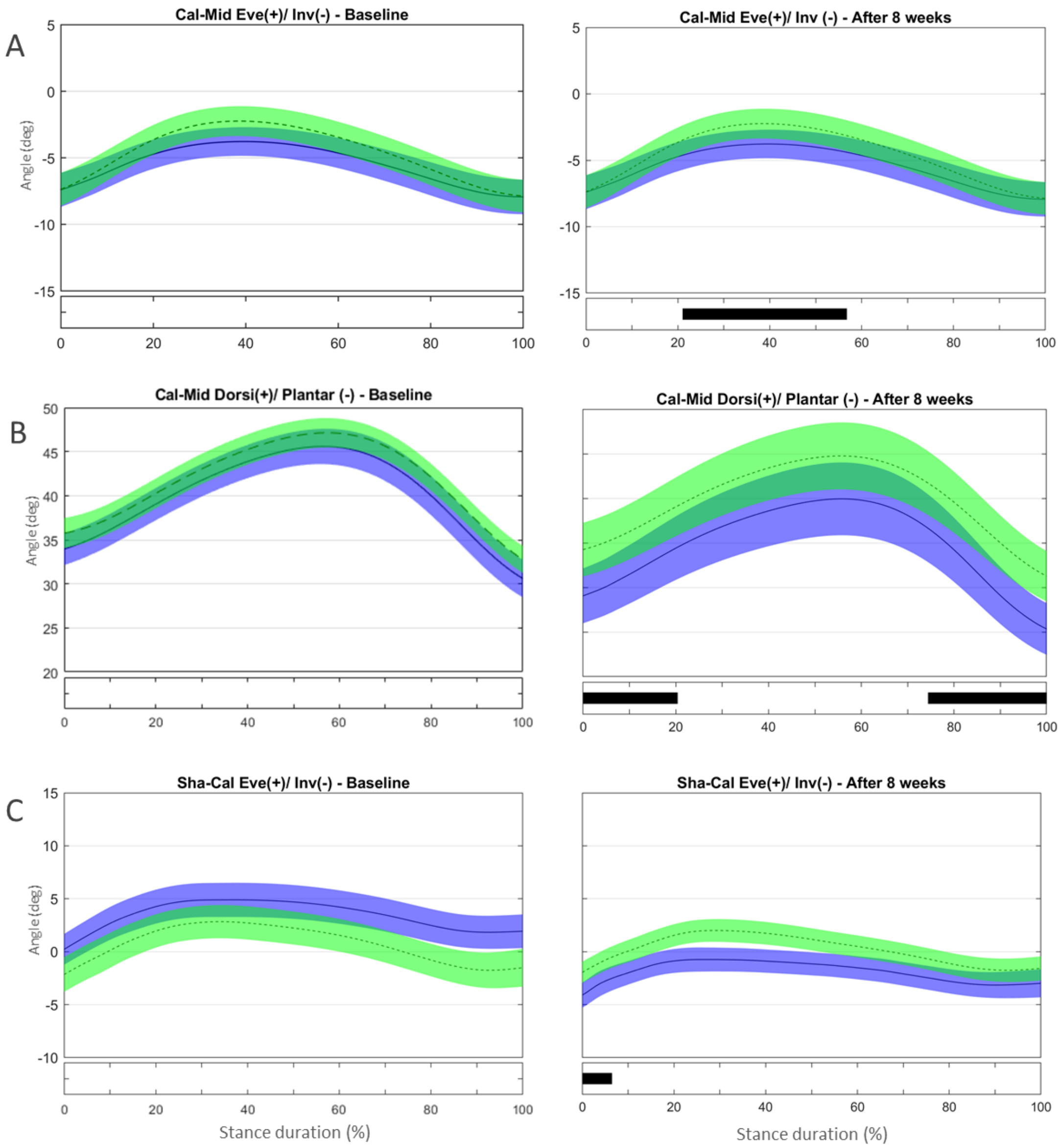

Figure 21 - Mean (±1SD) joint rotation angles during normalized stance phase duration of running in the CG (left) and IG (right). From top to bottom: sagittal-plane rotations between metatarsus and midfoot (A); sagittal-plane rotations between midfoot and calcaneus (B), and frontal-plane rotations between calcaneus and shank (C). Green, CG group; Blue, IG group. The black bar below the graph represents the time during which the differences between the groups occurred ($p<0.05$), what was indicated by the SPM{t} statistic.

At baseline, no difference was observed in the frontal-plane calcaneus to shank joint angle (Sha-Cal) between groups. At the 8-week assessment, calcaneus inversion at initial contact (0–6% stance, t*= 1.969, p=0.05) was greater in the IG compared to the CG (Figure 21C). This result is consistent with the outcome of the discrete analysis which showed a more inverted rearfoot in the IG. As stated before, a more inverted calcaneus at early stance may help attenuating the impact forces to the tibiotalar joint and diminishing tibia rotation[74], and thus may have contributed to the reduced RRI incidence in the IG[75].

At baseline, the IG showed a less adducted Sha-Cal than that in CG for most of stance duration (t*2.694, p<0.001), and no significant changes were observed after 8 weeks. Since both groups were different at baseline and maintained the differences after 8-weeks, the intervention does not seem to be responsible for the changes in the transverse Sha-Cal pattern observed in the IG after the training. No difference was found in sagittal-plane angle of Sha-Cal between groups after the intervention [see Additional file 2 – Figure 24].

Different from what we found in the discrete analysis; the SPM analysis did not reveal differences between groups in the MLA excursion during the whole stance phase after 8-weeks comparing CG and IG. Both groups presented more pronounced MLA angles with large intergroups ranges of motion at late stance after 8 weeks compared to baseline (89–100% stance; t* = 2.521, p = 0.025). In addition, a significant greater variability in the MLA pattern was present both in the IG and CG after 8 weeks [see Additional file 2 – Figure 24]. Both protocols may have affected MLA biomechanics, albeit in different directions, thus making it difficult to comprehend the differences between groups. It is notable that a stretching protocol designed as placebo intervention affected MLA kinematics in the CG. The choice for a simple stretching protocol was due to most participants being part of running groups which already had some sort of stretching routine. Since most participants would combine their running practice with muscle stretching, the CG adherence to the protocol was not controlled thoroughly. The stretching exercises resulted in stretching of the Achilles tendon and foot-ankle plantarflexors muscles that may directly influence the calcaneus inclination; this could in turn modify the tension in the plantar fascia and, consequently, the passive support of the MLA[76,77]. Thus, the potential changes in the MLA pattern expressed by the increased variability in CG may be consequence of the changes in plantar fascia tension and calcaneus position/motion.

However, another explanation may be considered. According to Pataky e t al. 2013, in some cases, scalar extraction analysis, based on the extraction of discrete variables that appear to have maximum effect, reaches significance and SPM analysis does not. Comparing discrete variables means, in fact, consider the comparison of only one sample of the entire time series, discarding the remaining sample by sample comparisons.

In summary, after the intervention, recreational runners landed with a more inverted calcaneus in relation to the shank, and a less dorsiflexed midfoot with respect to the calcaneus compared to controls. At midstance, runners presented a less plantarflexed metatarsus to the midfoot and more adducted to the calcaneus, and a more abducted 1st metatarso-phalangeal joint compared to controls. Last, the intervention resulted in a less dorsiflexed midfoot to the calcaneus at push-off, and in a less adducted and more dorsiflexed metatarso-phalangeal joint.

Strengths and limitations

The strengths of our study are: the rigorous method for the randomized controlled trial; its high completion and small drop-out rates at follow-up; the adoption of robust statistical models (GMM and 1D-SPM) that consider the complex non-linear iterations of foot joints biomechanics, and its large sample size compared to other studies in the same field[44,50,78–81].

This study has though some limitations that should be mentioned. First, we did not assess running biomechanics at 1-year follow-up as we did for RRI incidence; thus, we could not draw any conclusions about the causality between RRI incidence and kinetic- kinematic-related risk factors for RRI due to our training program. Secondly, while different RRIs or injury sites are expected to originate from different mechanisms and enhancing foot strength might be more effective in preventing some types of injuries than others, we could not differentiate between types of RRIs. This prevented us from explaining the biomechanical mechanisms for the reduction of RRI incidence in the IG after the intervention.

Furthermore, although lower limb kinematic patterns were shown to be similar between over-ground and treadmill running[82,83], the participants ran barefoot on a treadmill, a condition different from their usual practice. Finally, we observed some differences at baseline in the foot-ankle kinematic pattern which could be related to the previous identified clusters among our population of recreational runners[84]. After the foot-core training, runners in the different clusters might have responded differently to the program, thus suggesting that the response to the exercise intervention is dependent on the individual foot biomechanical pattern what could justify the absence of differences in some discrete and continuous outcomes analyzed.

Conclusion

The 8-week foot-core exercise program significantly changed the kinematic patterns of the ankle, tarso-metarsal, midtarsal and metatarso-phalangeal joints and some of the biomechanical risk-factor for RRI, such as MLA ROM and rearfoot angle. No effect was observed on impact and breaking forces or on loading rates. While a further mediation analysis should be sought, the observed changes in foot joint kinematics may be responsible for the reduction in RRI incidence following the foot-core training program.

Methods

Study design

A detailed protocol of the single-blind, two-arms parallel controlled randomized trial has been published elsewhere (Matias et al., 2016). The study was approved by the Ethics Committee of the School of Medicine of the University of São Paulo (18/03/2015, Protocol #031/15), and was registered with clinicaltrials.gov (Identifier NCT02306148) and a detailed flow chart of the study is described in the figure 22.

Participants

Adult recreational runners were recruited through digital social media advertising, posted flyers and direct contact with runners and running groups in the university surroundings recruited between August 2015 and August 2017. All participants were RRI-free, in the 2 months prior to baseline assessment, had no experience running barefoot or in minimalist shoes, were without chronic diseases or impairments that could influence running performance and participated in ≥ 1 year of running between 20 km and 100 km/week.

Sample size

We carried out a parallel group randomized controlled trial with 12 months' follow-up. In our previous study [51], an a priori sample size was calculated using several kinematic foot outcomes. Based on 80% power, and a significance level of 5%, the study indicated that we needed a total of 86 participants for most of the secondary outcomes. The V2G, S2G, F2G, S2F and S2V and the MLA required 38, 86, 58, 2184, 34 and 6 participants, respectively. An a posteriori effect size was also calculated for all variables presented and it is described in the results session.

Randomization and follow-up assessments

After the runners' agreement to participate and the application of the baseline questionnaire, participants were randomized into either the intervention (IG) or control group (CG), using the Clinstat software (University of York, Heslington, UK) to generate a randomization list with blocks of 8. The randomization list was developed by an individual who is not part of the research team. The codes for the groups were kept in opaque, sealed envelopes numbered from 1 to 120, and the researchers involved in the allocation and assessments were blind to the group codes and block size. The participants were enrolled and assigned to the interventions by a member of the research group.

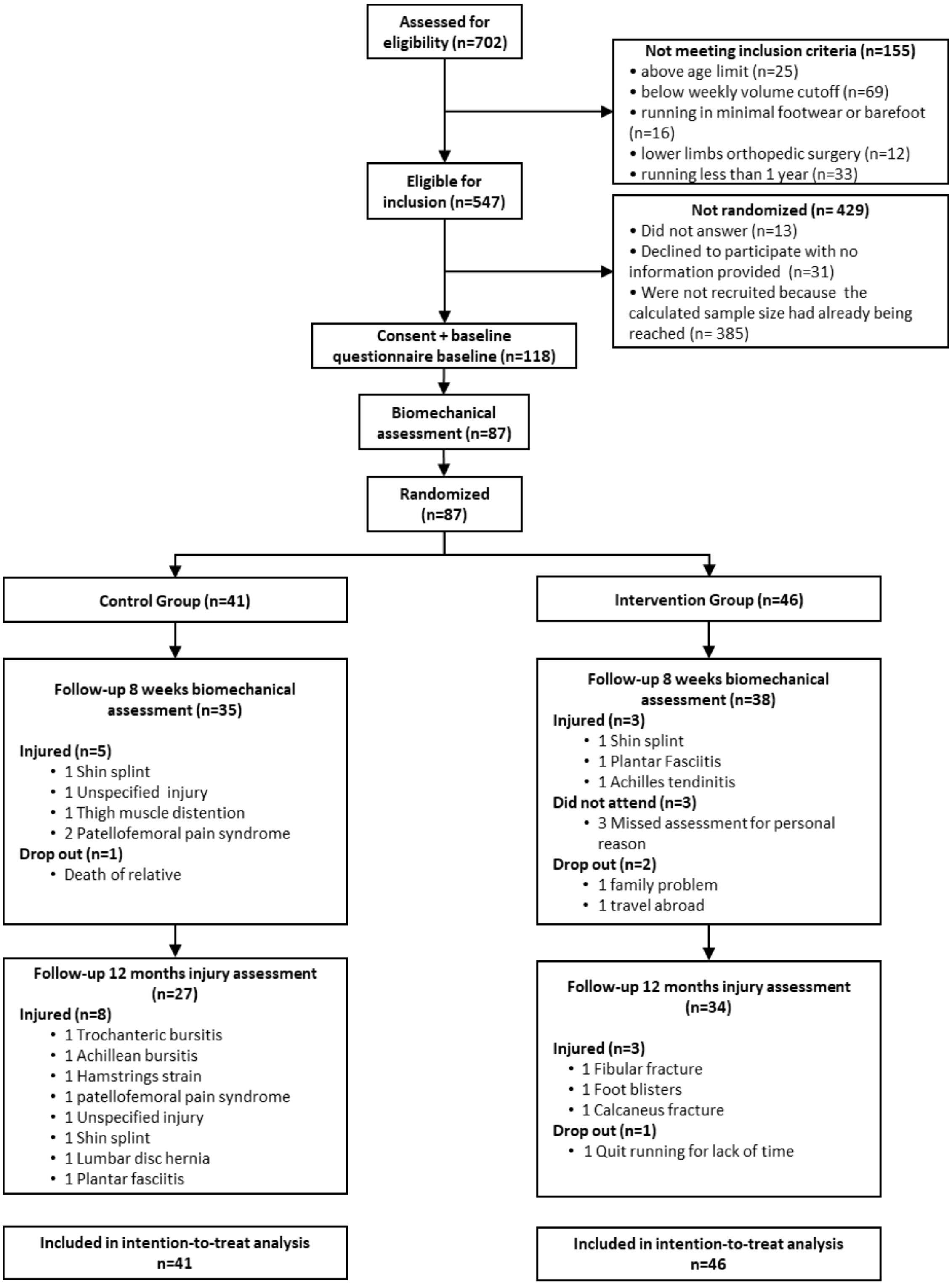

Figure 22 - Flowchart of recruitment, assessment, and follow-up process.

The trial statistician was blind to treatment allocation until the main analysis had been completed. All participants’ data were kept confidential before, during, and after the study by encoding their names. Participants allocated to the IG were given access to eight weeks of a training program. Participants in the CG were informed about their allocation into the control group and consequently were instructed to perform a 5-minute placebo static stretching protocol. We instructed the participants to keep their allocation group information strictly personal.

The baseline questionnaire consisted of six sections (demographics, training, running events, Foot Health Status Questionnaire and previous RRIs) (table 11). The follow-up questionnaires informed on running routine, on the foot core training program adherence and RRIs.

Table 11 - Baseline questionnaire and follow-up

Questionnaire	Section	Items
Baseline questionnaire	Demographics	Sex Age Body mass (kg) Height (cm) BMI (kg/m^2)
	Training	Running experience (years) Average running frequency over the last month (times per week) Average running distance over the last month (km/week) Average pace over the last month (min/ km)
	Running events	Member of athletic association (yes) Previous participation in running events (yes/no) Average participations in running events before
	FHSQ	Eight domains of the questionnaire
	Anthropometrics	Foot posture index Cavanagh Rodgers index
	Previous running-related injuries	Running-related injury in previous 12 months (yes/no) Location of running injury
Weekly follow-up questionnaires	Training	Running frequency (times/week) Running distance (km/week)
	Intervention protocol sessions	Number of foot exercise sessions completed
	New running-related injuries	New running-related injury since filling in previous questionnaire (yes/no) Location of new running-related injury Time to injury

Interventions

The foot-core training program to prevent RRI was developed focused on the foot-ankle muscles, with 12 exercises progressing weekly in volume and difficulty[49]. Participants in the IG were trained once a week by a physiotherapist and given online access to a web-based software developed for this project with the exercises' descriptions and videos to perform the same exercises an additional 3x/week, remotely supervised by the same physiotherapist.

For the control group, a 5-minute placebo static stretching protocol, to be performed 3x/week, based on online descriptions and images was also developed[49]. To improve adherence to the program, the participant was monitored using the same web-based software, with its importance reinforced at every contact with the subjects. Both groups were instructed to perform their respective exercises 3x/week up to the end of the 12-months follow up and to register their adherence in the web-software. The participants were strongly advised to not engage in any new exercise program during the intervention period.

Measurements

Biomechanical data were collected using an eight-camera motion caption system (Vicon Motion System Ltd., Oxford Metrics, UK) for the acquisition of 3D kinematic data at 200 Hz while running. The shank and foot were instrumented with 16 reflective skin-markers (9 mm diameter) according to the Rizzoli multi-segment foot model (RFM) [85,86]. Following a standing calibration trial, the participants were requested to run barefoot at a self-selected comfortable speed on an AMTI™ force-sensing tandem treadmill (AMTI, Watertown, MA) for the acquisition of ground reaction force data at 1000 Hz. In order to habituate to the treadmill and warm-up, the participants were instructed to run for 2–3 min before the data collection. A 30-s running trial was recorded at the self-selected comfortable speed after the accommodation period. Heel strike and toe off were identified when the vertical ground reaction force crossed a 30 N threshold. Kinematic and ground reaction force data were filtered using a fourth order, zero-lag, low-pass Butterworth filter with cut-off frequencies of 10 and 80 Hz, respectively. The outputs of the RFM were calculated by custom-made scripts in Visual3D (Visual3D, C-Motion, German- town, MD) according to the published definitions[85–87]. Joint rotations were calculated using the Joint Coordinate System[88] convention. The axes of each joint reference frame were defined as follows: sagittal plane rotations around z-axis (medio-lateral); frontal plane rotations around x-axis (anterior-posterior); and transverse plane rotations around y-axis (vertical). Data were normalized to 0–100% of stance phase.

Outcomes

This study is an analysis of the secondary outcomes from the developed RCT. The primary outcome variable was incidence of RRI in recreational runners over the course of a one-year follow up and was published elsewhere[52]. The secondary outcomes were related to foot-ankle kinematics and running kinetics. The foot kinematic variables were as follows: 3D MLA[87] excursion (max-min)**;** rotation angles in the three anatomical planes between shank and calcaneus (Sha-Cal), calcaneus and midfoot (Cal-Mid), midfoot and metatarsus (Mid-Met), calcaneus and metatarsus (Cal-Met), and first metatarsus and hallux (Met-Hal). The following metatarsal bone angles were also assessed: sagittal-plane inclination of the 1st metatarsal bone to the ground (F2G), 2nd metatarsal bone to the ground (S2G), and 5th metatarsal bone to the ground (V2G); and transverse-plane divergence between 1st and 2nd metatarsal bones (S2F), and between 5th and 2nd metatarsal bones (S2V). In addition, kinematic and kinetic biomechanical-related risk factors for RRI were investigated as discrete parameters: rearfoot angle (Sha-Cal frontal angle peaks), MLA ROM (range of motion: max-min), vertical average loading rate (vertical force rate between 20 and 80% between the foot contact and the first peak), horizontal breaking forces (maximum posterior force, horizontal component), and vertical impact peak (local maximum vertical force at initial contact).

Statistical analyses

All analyses used the full set of randomly assigned participants under the intention-to-treat assumption. Generalized Linear Mixed Model (GLMM) method was used for univariate analyses, considering the following as factors: groups (CG and IG), time of assessment (baseline and after 8 weeks), and the interaction effect (time by group), which was our primary outcome comparison. Participants and time were considered as random effects and groups as fixed effects in the GLMM modeling. Q-Q graphs were plotted to verify the adequacy (normality) of each model. Univariate (main and interaction effects) comparisons of the estimated marginal means were adjusted with the Bonferroni correction. The comparisons between the pairs of estimated marginal means were made based on the original scale of each of the dependent variables of the study. Statistical analyses were performed using the *Statistical Package for the Social Science* (SPSS, IBM; v.26.0), adopting a 5% significance level.

Additionally, to capture features of the entire time series, a vector field analysis of the resultant angles was conducted using the 1-D statistical parametric mapping (1-D SPM), as described elsewhere[89,90]. A custom-written MATLAB code (MATLAB 2020a, MathWorks, Natick, USA), using the source code available at http://www.spm1d.org/, was employed in the analysis. The SPM captures features of the entire time series, rather than a few discrete variables, and can provide additional information. Each component of each time series was interpolated to contain 101 points (0-100% of the stance phase) and organized in an array with two or three corresponding matrices, one for each variable component, 87 rows, one for each subject, and 101 columns. 1D SPM ANOVA followed by post hoc SPM t-tests was used for 1-dimensional variables (F2G, S2G, V2G, S2F, S2V, and MLA). Paired (for assessment comparisons) and independent (for group comparisons) Hotelling's T2-test were used for 3D variables comparisons (Sha-Cal, Cal-Mid, Mid-Met, Cal-Met, and Met-Hal) in a 3D vector field SPM analysis, followed by the paired or independent t-test, as a post-hoc test, with a Sidák correction. The output of SPM provides T^2, F, and t values for each sample of the investigated kinematic time series, and the threshold corresponding to the set alpha level [see Additional file 3]. The T^2, F, and t values exceeding this threshold (marked as black bars below each figure, e.g. in Fig. 20) indicate significant differences in the corresponding portion of the time series.

Author contributions

AM, UT and ICNS were responsible for the conception and design of the study; AM, RW, UT, RS, RB were responsible for data acquisition and data processing; PC and MF for developing tools for the analysis; AM, RW, UT, PC, RS, RB, YS, MF and ICNS for data

analysis and interpretation; and AM and ICNS for drafting the paper. ICNS revised the manuscript critically. All authors read, provided feedback and approved the submitted version.

Compreting interests

None of the authors had any financial or personal conflict of interest with regard to this study.

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Availability of data and materials

The raw data supporting the conclusions of this manuscript can be accessed through the following data repository link: https://doi.org/10.6084/m9.figshare.13484715.v2

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CAPÍTULO VI -
CONSIDERAÇÕES FINAIS

6.1 Discussão geral

Esta tese teve como objetivo investigar aspectos biomecânicos do pé na corrida, bem como investigar, por meio de um ensaio clínico randomizado e controlado, a eficácia de uma intervenção fisioterapêutica inovadora focada na musculatura dos pés, na cinemática do pé e nas forças e taxas de carga durante a corrida em corredores fundistas recreacionais. Antes e durante a execução do ensaio clínico principal, desenvolvemos estudos complementares para melhor compreender a biomecânica do pé em função de diferentes tipos de aterrissagens na corrida e investigar a usabilidade e confiabilidade das medidas do modelo multisegmentar do pé que foram os desfechos principais do ensaio clínico. A seguir, apresentaremos uma síntese dos principais achados desta tese, e discutiremos as implicações para a prática da corrida, as implicações clínicas e as considerações para estudos futuros.

A primeira etapa desta tese foi propor um protocolo de exercícios para os pés e tornozelos para corredores recreacionais saudáveis de longa distância que apresentam uma alta prevalência de lesões musculoesqueléticas (1). Descrevemos em detalhes todo o procedimento metodológico do ensaio clínico randomizado de grupo paralelo, simples-cego e controlado que incluiu 118 corredores recreacionais alocados aleatoriamente, para um grupo que executou alongamentos de membros inferiores dentro de sua rotina da corrida (grupo controle) ou para um grupo que executou exercícios terapêuticos para os pés e tornozelos supervisionados, presencialmente, uma vez por semana e, remotamente, 3 vezes por semana (grupo intervenção), durante 8 semanas (2). Os participantes foram acompanhados por um período de 1 ano em relação aos desfechos primários (ocorrência de lesão relacionada à corrida) e avaliados 3 vezes em relação às variáveis secundárias: comportamento dinâmico do arco longitudinal plantar, cinemática de tornozelo e pé, forças e taxas de carga durante a corrida, bem como saúde e funcionalidade do pé e força muscular do pé.

Já se sabe que o pé desempenha várias funções importantes durante a corrida, tais como receber e atenuar cargas, adaptar-se ao solo e armazenar e devolver energia elástica (3,4). Estudos longitudinais já mostraram que intervenções cinesioterapêuticas simples podem ser aplicadas no pé e até mesmo jovens saudáveis beneficiam-se dessas intervenções (5–7). Assim, essa abordagem chamada de "bottom-up", com foco nas articulações de tornozelo-pé, poderia ter efeitos positivos na biomecânica das articulações de tornozelo, joelho e quadril e, até mesmo, reduzir lesões em corredores (8–11).

Diante disso, o protocolo publicado (artigo 2.1 do capítulo II) descreveu um ensaio clínico randomizado e controlado para avaliar os efeitos dessa intervenção inovadora para o tornozelo e pé, com um período longo de acompanhamento (12 meses), incluindo desfechos relacionados à biomecânica da corrida e um tamanho amostral calculado com poder suficiente para responder aos desfechos escolhidos (2).

Para avaliar os efeitos do programa de exercícios terapêuticos no comportamento biomecânico do pé durante a corrida, utilizamos um modelo cinemático mutisegmentar desenvolvido no Instituto Ortopédico Rizzoli (Bolonha, Itália). Os modelos cinemáticos multisegmentares de pé baseados em marcadores de pele estão crescendo em popularidade para a avaliação dos segmentos do pé em diferentes tarefas motoras. No entanto, ainda é escassa a descrição dos efeitos de tarefas motoras de alto impacto e velocidade, como a corrida, na repetibilidade dessas medidas. Logo, a segunda etapa desta tese buscou avaliar a confiabilidade, a usabilidade e a acurácia das medidas do modelo multisegmentar de pé Rizzoli na corrida, que seriam usadas como desfechos do ensaio clínico. O primeiro estudo dessa etapa buscou avaliar e comparar a repetibilidade Intertentativas, intersessões e interavaliadores de medidas cinemáticas dos segmentos do pé durante o andar e o correr em adultos saudáveis. Esse estudo resultou no artigo 3.1 do capítulo III e concluiu que o modelo de pé Rizzoli é confiável para ser utilizado na corrida, muito embora sua confiabilidade interexaminadores seja mais baixa em relação a sua confiabilidade no andar.

Um dos desfechos importantes do ensaio clínico foi a deformação do arco longitudinal medial do pé durante a corrida. O arco foi escolhido devido ao protagonismo do pé no mecanismo de armazenamento e retorno de energia elástica por meio de sua deformação quando sob cargas, além de seu papel fundamental na atenuação e distribuição de cargas durante tarefas locomotoras. Os modelos de avaliação cinemática do arco longitudinal permitem estimar o seu formato e sua deformação, mas, aparentemente, as marcas usadas para a configuração do ângulo do arco não são consistentes com a anatomia do pé e com as definições clínicas padrões, tais como as usadas em radiografias. No artigo 3.2 do capítulo III, propusemos outras configurações de marcas que se aproximavam mais da anatomia do arco, além de testarmos e compararmos a reprodutibilidade inter e intraexaminadores e intersessões de oito possíveis definições do arco. Foram usadas variações com marcas no calcâneo, na tuberosidade do navicular, no sustentáculo do tálus, na cabeça e na base do primeiro metatarso e nos maléolos lateral e medial, projetadas tanto bi quanto tridimensionalmente. Os resultados de nossos estudos mostraram que o modelo cinemático do arco original do Rizzoli foi o que apresentou a melhor reprodutibilidade quando projetado tridimensionalmente, pois sua variabilidade interexaminadores e intersessões foi a menor dentre todas as variações propostas testadas. A configuração original utiliza a marca do calcâneo, do sustentáculo do tálus e da cabeça do primeiro metatarso. O ângulo do arco nessa configuração é calculado utilizando uma reta traçada entre a projeção da marca do calcâneo no chão e o sustentáculo do tálus e uma outra reta traçada entre o sustentáculo do tálus e a cabeça do primeiro metatarso. O ângulo formado entre essas duas retas distintas e concorrentes projetado no plano sagital do pé é considerado o ângulo que representa o arco longitudinal medial.

Ainda com o objetivo de aprimorar o modelo cinemático multisegmentar do pé que usaríamos no ensaio clínico, avaliamos a correlação e a acurácia das medidas cinemáticas de deformação do arco em relação às medidas radiográficas clínicas padrões quando o pé é deformado (artigo 3.3 do capítulo III). Observamos que a deformação do arco é altamente afetada pelo modelo de marcas escolhido. O novo modelo de arco desenvolvido por nós, utilizando a tuberosidade do navicular como vértice do arco, é o que demonstrou ser mais acurado quando comparado às medidas radiográficas e quando comparado ao modelo original. Assim, embora o novo modelo proposto não tenha obtido a melhor reprodutibilidade, foi o que gerou medidas mais similares à anatomia e à medida radiográfica usada rotineiramente na clínica. Concluímos que as definições do ângulo do arco, com base em conjuntos que usam um número mínimo de marcadores, são as mais recomendadas para uma maior confiabilidade das medições durante as coletas, porém usando a tuberosidade do navicular como vértice do arco obtém-se uma correspondência mais fiel à anatomia e às medidas clínicas.

Na terceira etapa da tese, estudamos como o tipo de aterrissagem na corrida (antepé ou retropé) influenciava a cinemática dos segmentos pé e das forças e taxas da força vertical durante a corrida. No artigo 4.1 do capítulo IV, concluímos que a forma como o pé entra em contato com o solo na corrida determina diretamente como será o comportamento cinemático do restante dos segmentos do pé na fase de apoio. Em particular, corredores de antepé aterrissam com os metatarsos mais próximos ao solo, mantêm os metatarsos menos aduzidos e, após a aterrisagem, mantêm o tornozelo mais dorsifletido durante toda a fase de apoio, retornando à flexão plantar na fase de propulsão. Ainda, os corredores de antepé apresentaram o calcâneo mais aduzido em relação à perna e o mediopé mais invertido em relação ao calcâneo durante toda a fase de apoio. Também concluímos que não era simplesmente a forma de aterrissagem que determinava o padrão do impacto na corrida, mas se a altura do calcanhar em relação ao solo no contato inicial fosse menor, se o tempo de chegada do calcanhar ao solo fosse mais curto e se a aceleração do toque do calcanhar no contato inicial fosse maior. Nesta configuração dinâmica da corrida, as taxas de carga e os picos de impacto vertical eram maiores, mesmo em corredores de antepé (resultados artigo 4.2 do capítulo IV).

Na terceira e última etapa desta tese concluímos o ensaio clínico que implementou o programa de exercícios específicos para o fortalecimento dos músculos intrínsecos e extrínsecos do tornozelo e pé em corredores recreacionais fundistas. Esta intervenção fisioterapêutica proposta foi capaz de modificar os padrões cinemáticos do pé e de fatores de risco biomecânicos para lesões no grupo intervenção comparado ao grupo controle. Houve uma diminuição da amplitude de movimento do arco longitudinal medial e um aumento da inversão do calcâneo. Ainda, depois da intervenção, os corredores passaram a aterrissar com o mediopé menos dorsifletido em relação ao calcâneo, com o calcâneo mais invertido em relação à perna e com o tornozelo menos invertido comparado ao grupo controle. No médio apoio, os participantes do grupo intervenção correram com o hálux mais abduzido em relação aos metatarsos, com menor flexão plantar do metatarso em relação ao mediopé e mais adução em relação ao calcâneo, e com o tornozelo menos aduzido comparado ao grupo controle. Na fase de propulsão da corrida, a intervenção resultou no mediopé menos dorsifletido em relação ao calcâneo e o hálux menos aduzido e mais dorsifletido em relação ao metatarso.

Os resultados dos estudos apresentados nesta tese trazem algumas contribuições clínicas para a prevenção de lesões e para a reabilitação de corredores, embora não tenha havido efeito nas forças de impacto e na taxa de carga durante a corrida, que são fatores de risco biomecânicos relativamente estabelecidos (12–17).

A primeira contribuição é o conhecimento do comportamento dos segmentos do pé em diferentes aterrisagens na corrida. Nos estudos que compõem esta tese, por termos o foco no pé, a corrida foi avaliada descalça, permitindo a melhor descrição dos movimentos de cada segmento do pé, diferente de estudo anteriores que, em geral, realizaram a corrida calçada. Cada tipo de aterrissagem (antepé ou retropé) resultou em uma resposta diferente nos segmentos do pé, podendo aumentar ou diminuir as forças atuantes nos tecidos moles que apresentam diferentes riscos de lesão (18). Ainda, caso o corredor opte por trocar o padrão de pisada com o objetivo de reduzir a ocorrência de lesões de repetição, os resultados do estudo mostram que a corrida de antepé não garante a eliminação do pico de impacto e pode apresentar altas taxas de carga similares à corrida de retropé. Os achados podem ser de interesse para médicos, fisioterapeutas e outros profissionais de saúde para apoiar estratégias de prevenção ou reabilitação de tipos específicos de lesões associadas aos diferentes tipos de pisada, como fascite plantar, tendinopatia do tendão calcâneo e fraturas por estresse de metatarsos, por meio de orientações biomecânicas específicas quanto aos padrões de pisada (19).

Outra contribuição clínica da tese foi demonstrar que o fortalecimento da musculatura intrínseca dos pés foi capaz de reduzir a amplitude de movimento do arco longitudinal durante a corrida e, provavelmente, esse ganho de sustentação do arco aumentou a resistência do mesmo ao colapso, o que ocorre principalmente na fase de contato inicial e médio-apoio na corrida. Um arco mais resistente ao colapso/deformação durante os esforços repetitivos da corrida deve diminuir as demandas mecânicas sobre os tecidos moles do pé, como tendões, fáscia e ligamentos, potencialmente reduzindo o risco de lesões nessas estruturas (18,20). Ainda em relação ao programa de fortalecimento da musculatura do complexo tornozelo-pé, a tese demonstrou que estes exercícios específicos podem ter contribuído para o aumento da força e funcionalidade do músculo tibial posterior que, provavelmente, foi o responsável pelo melhor controle da eversão do calcâneo na fase de apoio. Um calcâneo mais invertido, observado após a intervenção, poderia aumentar a torsão da placa osteoligamentar no apoio inicial na corrida e, consequentemente, resistir à pronação durante a fase de apoio, quando a placa tende a distorcer. Ao resistir à pronação, a articulação tibiotalar deve ficar mais protegida de altas forças de tração no apoio (21), diminuindo as chances de lesão relacionadas a corrida.

Cabe destacar, ainda, que o protocolo de exercícios proposto nesta tese, além de efetivo para a redução de lesões relacionadas à corrida (22) e para modificar positivamente a biomecânica do pé, é prático, fácil e seguro de ser implementado de forma mais ampla na rotina de corredores recreacionais. A diminuição da incidência de lesões em corredores recreacionais poderia diminuir o tempo de afastamento dos treinos, o consumo de medicamentos e, até mesmo, as faltas ao trabalho causadas pelas lesões relacionadas à corrida (23).

Por fim, além das contribuições clínicas da tese mencionadas, os resultados, principalmente do capítulo III, devem contribuir com futuros estudos que pretendem avaliar a cinemática dos segmentos do pé na corrida.

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ANEXO 1 – APROVAÇÃO DA COMISSÃO DE ÉTICA DA FACULDADE DE MEDICINA DA UNIVERSIDADE DE SÃO PAULO PARA PROJETO DE PESQUISA

MEDICINA
USP
COMITÊ DE ÉTICA EM PESQUISA

APROVAÇÃO

O Comitê de Ética em Pesquisa da Faculdade de Medicina da Universidade de São Paulo, em sessão de 18/03/2015, **APROVOU** o Protocolo de Pesquisa nº **031/15** intitulado: **"EFEITOS DO TREINAMENTO DA MUSCULATURA DO PÉ NA PREVALÊNCIA DE LESÕES EM CORREDORES FUNDISTAS: UM ENSAIO CLÍNICO CONTROLADO E RANDOMIZADO"** apresentado pelo **Departamento de FISIOTERAPIA, FONOAUDIOLOGIA E TERAPIA OCUPACIONAL**

Cabe ao pesquisador elaborar e apresentar ao CEP-FMUSP, os relatórios parciais e final sobre a pesquisa (Resolução do Conselho Nacional de Saúde nº 466/12, inciso IX.2, letra "c").

Pesquisador (a) Responsável: Isabel de Camargo Neves Sacco
Pesquisador (a) Executante: Ulisses Taddei Tirollo
Pesquisador (a) Executante: Alessandra Bento Matias

CEP-FMUSP, 18 de Março de 2015.

Prof. Dr. Roger Chammas
Coordenador
Comitê de Ética em Pesquisa

Comitê de Ética em Pesquisa da Faculdade de Medicina
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ANEXO 2 – TERMO DE CONSENTIMENTO LIVRE E ESCLARECIDO

Projeto de pesquisa: ***"Efeitos do treinamento da musculatura do pé na prevalência de lesões em corredores fundistas: um ensaio clínico controlado e randomizado".***

Eu,____, concordo em participar da pesquisa conduzida pela Profa. Dra. Isabel de Camargo Neves Sacco, pelo Prof. Dr. Marcos Duarte, pelo Fisioterapeuta Ulisses Tirollo Taddei e pela Fisioterapeuta Alessandra Bento Matias do Laboratório de Biomecânica do Movimento e Postura Humana do Departamento de Fisioterapia, Fonoaudiologia e Terapia Ocupacional, da Faculdade de Medicina, da Universidade de São Paulo. Os resultados, guardadas as devidas identificações e mantida a confidencialidade, serão analisados e utilizados única e exclusivamente para fins científicos. Este projeto tem como objetivo **investigar os efeitos do treinamento da musculatura dos pés na prevenção de lesões em corredores.**

Explicação dos procedimentos:

- Etapa 1:

Esta etapa ocorrerá no ocorrerá no Laboratório de Biomecânica do Movimento e Postura Humana da USP, localizado na Cidade Universitária e conta com dois questionários, uma avaliação da sua corrida e da força dos seus pés. O primeiro questionário identifica seu nome, idade, altura, peso, telefone para contato, endereço, tipos de medicamentos que o(a) senhor(a) usa, tipo de tênis e detalhes do seu programa de treinamento de corrida (volume, frequência), além do histórico de lesões. O segundo questionário avaliará o estado de saúde dos seus pés. Para a avaliação da corrida, colocaremos marcadores (bolinhas prateadas de isopor) em determinados pontos do seu corpo e o(a) senhor(a) caminhará algumas vezes pelo laboratório. A força dos seus pés será avaliada com o(a) senhor(a) sentado numa cadeira movimentando seus pés, tornozelos e joelhos e de pé pisando sobre uma plataforma.

Por fim, lhe informaremos se o(a) senhor(a) fará parte do Grupo que receberá um treinamento para fortalecimento dos pés a partir de agora ou ao término do estudo e se fará parte do grupo que realizará o exame de ressonância magnética.

- Etapa 2:

O treinamento para fortalecimento dos pés terá duração de 8 semanas, contando com exercícios realizados presencialmente e não supervisionadas. Após as 8 semanas de intervenção presencialmente supervisionada, os corredores continuarão a prática dos exercícios de forma independente e não remotamente supervisionada em domicílio utilizando o mesmo software com progressões programadas individualmente, 3 vezes por semana até o final do estudo (por mais 12 meses). O uso do software em domicílio durante os primeiros 2 e 12 meses seguintes ao início da intervenção será monitorada pelo seu acesso ao software e, também, segundo o preenchimento dos formulários de realização dos exercícios. O preenchimento deverá ser realizado semanalmente nos primeiros 6 meses do início do estudo e quinzenalmente a partir de então até o término do estudo.

- Etapa 3: O(a) senhor(a) deverá retornar ao laboratório de biomecânica do departamento de Fisioterapia (Cidade Universitária – USP) após 2 e 6 meses da data de início do estudo para reavaliarmos reavaliação da sua força, da sua corrida e aplicação dos mesmos questionários da primeira visita.

Desconforto e risco: o experimento não envolverá qualquer desconforto ou risco à sua saúde física e mental, além dos riscos encontrados nas atividades normais que o(a) senhor(a) realiza diariamente.

Benefícios: Caso o/a senhor(a) seja sorteado para o grupo de exercícios, o(a) senhor(a) receberá gratuitamente um programa para fortalecimento dos pés durante 12 meses remotamente, sendo presencialmente supervisionada durante 8 semanas (2 vezes por semana). Caso o(a) senhor(a) seja sorteada para o grupo controle (sem os exercícios de fortalecimento), o(a) senhor(a) irá receber um treinamento para fortalecimento dos pés. Além de receber o programa de fortalecimento, o(a) senhor(a) irá contribuir para o entendimento da importância dos pés nas prevenções de lesões em corredores.

Garantia de acesso: Em qualquer etapa do estudo você terá acesso aos profissionais responsáveis pela pesquisa para esclarecimento de eventuais dúvidas. O principal investigador a profª. Drª Isabel de Camargo Neves Sacco que pode ser encontrado no Laboratório de Biomecânica do Movimento e Postura Humana, Departamento de Fisioterapia, Fonoaudiologia e Terapia Ocupacional, na rua Cipotânea, 51, Cidade Universitária (telefone 3091-9426) Se você tiver alguma consideração ou dúvida sobre a ética da pesquisa, entre em contato com o Comitê de Ética em Pesquisa (CEP) – Rua Ovídio Pires de Campos, 225 – 5º andar – tel: 3069-6442 ramais 16, 17, 18 ou 20, FAX: 3069-6442 ramal 26 – E-mail: cappesq@hcnet.usp.br

É garantida a liberdade da retirada de consentimento a qualquer momento e deixar de participar do estudo, sem qualquer prejuízo à continuidade de seu tratamento na Instituição. É seu direito ser mantido atualizado sobre os resultados parciais das pesquisas, quando em estudos abertos, ou de resultados que sejam do conhecimento dos pesquisadores.

Despesas e compensações: não há despesas pessoais para o participante em qualquer fase do estudo, incluindo exames e consultas. Também não há compensação financeira relacionada à sua participação. Se existir qualquer despesa adicional, ela será absorvida pelo orçamento da pesquisa. Os resultados verificados serão guardados com suas devidas identificações e mantidos em confidencialidade, os quais serão utilizados única exclusivamente para fins científicos.

Acredito ter sido suficientemente informado a respeito das informações que li ou que foram lidas para mim, descrevendo o estudo sobre **a investigar os efeitos do treinamento da musculatura dos pés na prevenção de lesões em corredores.**

Eu discuti com os responsáveis: Profª Drª. Isabel de Camargo Neves Sacco e/ou Ft. Ulisses Taddei e Ft. Alessandra Bento Matias sobre a minha decisão em participar nesse estudo. Ficaram claros para mim quais são os propósitos do estudo, os procedimentos a serem realizados, seus desconfortos e riscos, as garantias de confidencialidade e de esclarecimentos permanentes.

Ficou claro também que minha participação é isenta de despesas e que tenho garantia do acesso a tratamento hospitalar quando necessário. Concordo voluntariamente em participar deste estudo e poderei retirar o meu consentimento a qualquer momento, antes ou durante o mesmo, sem penalidades ou prejuízo ou perda de qualquer benefício que eu possa ter adquirido, ou no meu atendimento neste Serviço.

____

Assinatura do paciente/representante legal — Data ___/___/____

____

Assinatura da testemunha — Data ___/___/____

(Somente para o responsável do projeto)

Declaro que obtive de forma apropriada e voluntária o Consentimento Livre e Esclarecido deste paciente ou representante legal para a participação neste estudo.

____

ANEXO 3 – APROVAÇÃO DO COMITÊ DE ÉTICA (IRB) DO MASSACHUSETTS GENERAL HOSPITAL PARA PROJETO DE PESQUISA

HUMANS

- Action Required 0
- Notifications 12

2012P002373

- Forms
- Staff
- Attachments
- Related Records 1
- Summary Notes
- Actions
  - Create Protocol
  - Create CEDE Review
  - Create Amendment
  - Create Staff Updates
  - Create Other Event
  - Create Continuing Review or Continuing Review/Amendment
- Agreements
- Animals
- Biosafety
- Data Management
- Disclosures
- Manage Profiles & Security

Investigation of Running Mechanics in Healthy Runners

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Currently approved (v52.0)

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Protocol #: 2012P002373 | Overall Status: Active, Enrolling Subjects | Expiration Date: 03/18/22

PI, Institution: Davis, Irene S, SRH | Type: Intervention/Interaction | Protocol Admin: Papin Joseph, Line J

Staff

Search Study Staff History

Study Staff (5)

Name	Degree	Organization	Role	Permission	Process	DCR	eCOI	CITI Expires	Staff Cert.	Contact
Bento Matias, Alessandra		SRH > Dept of Physical Med and Rehab	Research Assistant	View	AME38			Not Certified	PDF	
Davis, Irene Sprague		SRH > Dept of Physical Med and Rehab	Principal Investigator	Manage	IR		In Review	07/23/24	PDF	
Fokkema, Tryntsje		SRH	Research Assistant	View	AME34			Not Certified	PDF	
Johnson, Caleb D.	D.PHIL, MS, BS	SRH	Research Assistant	Manage	SU1		Review Not Needed	07/09/24	PDF	
Outerleys, Jereme	MSC, BS, BS	SRH	Data Coordinator/Manager	Manage	AME32		Review Not Needed	01/12/24	PDF	

Non-study Staff (0)

Name	Organization	Permission	Comments	Contact

No data found.

Financial Delegate

Fund #	Financial delegate

No data found.

< Previous

Next >

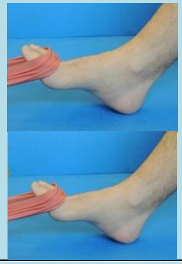
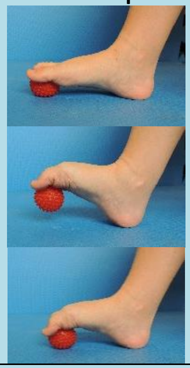
ANEXO 4 – Additional file 1 do Capítulo II

Table S1 - Exercises included in the supervised sessions by a physiotherapist.

Name	Execution	Training Volume	Progression	Progression Parameter	Approximate Duration
Massage 	Sitting, with leg crossed over the other, massage the sole of your feet with both hands, for 20 seconds. Rub your foot in a circular motion using your thumb. Do the same on the other foot.	1 set of 20 seconds each foot	-	-	40 Seconds
Toes manipulation 	Sitting, with leg crossed over the other, hold each toe and slowly spin side to side, like a screw. Do with all toes.	1 set of 10 times each finger	-	-	1 minute

Rubber ball slide 	Slowly slide your foot on the ball throughout the foot sole from the heel to the fingertips.	1 set of 30 seconds each foot	-	-	1 minute
Feet tapping 	With the heel fixed, tap your foot as fast as possible. Starts seated on a chair, and do with both feet at the same time. After you learn, do the same tapping standing.	1 set of 30 repetitions	1: 1x30 repetitions; 2: 2x30 repetitions; 3: 2x40 repetitions	Being able to perform the set without pain or muscle cramp after the completion of the set.	1-2 minutes
Forefoot ascend 	Standing, ascend and descend on forefoot. Start standing, using both feet. Use a chair or table to keep balance.	1 set of 30 repetitions	1: 1x30 repetitions; 2: 2x30 repetitions; 3: 2x40 repetitions	Being able to perform the set without pain or muscle cramp after the completion of the set.	1-2 minutes

Invert/Evert asymmetric 	Sitting, with 90 degrees of knee and ankle flexion, perform asymmetrical foot inversion (lifting medial side) and eversion (lifting lateral side).	1 set of 10 repetitions maintaining each position for 1 second.	1: Sitting: 1x10 repetitions; 2: Standing: 1x10 repetitions ; 3: Standing 1x20 repetitions maintaining each position for 2 seconds.	Being able to perform the set without pain or muscle cramp after the completion of the set, and without loss of balance.	1-2 minutes
Foot abduction 	Standing, using a resistance band around the forefoot, perform foot abduction and return to the original position	2 sets of 10 repetitions each foot	1: 2x10 repetitions; 2: 4x10 repetitions; 3: 6x10 repetitions.	Being able to perform the set without pain or muscle cramp after the completion of the set.	1-6 minutes

Toes and ankle flexion 	Sitting posture, using a resistance band around the forefoot, perform ankle and toes flexion and return to the original position	1 set of 10 repetitions each foot	1: 1x10 repetitions; 2: 2x10 repetitions; 3: 3x10 repetitions.	Being able to perform the set without pain or muscle cramp after the completion of the set.	1-3 minutes
Grab and hold squeeze ball 	Grab and hold a squeeze ball with all the toes, raise it from the floor and place it back to it's original position. Always keep the heel fixed on the ground.	1 set of 5 repetitions each foot holding the ball for 5 seconds	1: Sitting posture 1x5 repetitions; 2: Standing posture 2x5 repetitions; 3: Standing posture 3x5 repetitions.	Being able to perform the set without pain or muscle cramp after the completion of the set.	2-6 minutes
Squeeze toes separators 	Sitting position, with 90 degrees of knee and ankle flexion, adduct and abduct, squeeze the toes separators for one second always keeping the heel fixed on the ground.	1 set of 10 repetitions each foot	1: 1x10 repetitions; 2: 2x10 repetitions; 3: 3x10 repetitions.	Being able to perform the set without pain or muscle cramp after the completion of the set.	2-6 minutes

Squeeze ball with little toes 	Grab and hold a squeeze ball with the metatarsophalangeal region and place it back to the starting position.	1 set of 5 repetitions each foot holding the ball for 5 seconds	Progression requires raising squeeze balls hardness.	Being able to perform the set without pain or muscle cramp after the completion of the set and being able to hold abduction for the stipulated time.	2 minutes
Toes Abduction/adduction 	Sitting position, with 90 degrees of knee and ankle flexion, adduct and abduct toes holding each position for 2 seconds.	1 sets of 10 repetitions each foot holding abduction for 2 seconds and adduction for 2 seconds.	1: Sitting posture 1x10 repetitions; 2: Standing posture 2x10 repetitions; 3: Standing posture 2x10 repetitions holding abduction/abduction for 5 seconds.	Being able to perform the set without pain or muscle cramp after the completion of the set and being able to hold abduction for the stipulated time.	1-2 minutes

Short-foot exercise 	Sitting, with 90 degrees of knee and ankle flexion, approximate the head of the first metatarsal toward the heel without toe flexion, “shortening” the feet. The forefoot and heel should not get off the ground.	1 set of 10 repetitions each foot, maintaining 5 seconds each contraction.	1: Sitting 1x10 repetitions; 2: Standing 1x10 repetitions; 3: Single leg stance 1x10 repetitions.	Being able to perform the set without pain or muscle cramp after the completion of the set.	4-6 minutes
Plantar arch raise 	Sitting, raise the plantar arch in an arch shape. The heel and fingertips should not get off the ground.	1 set of 10 repetitions each foot, maintaining 5 seconds each contraction.	1: Sitting 1x10 repetitions; 2: Standing 1x10 repetitions; 3: Single leg stance 1x10 repetitions.	Being able to perform the set without pain or muscle cramp after the completion of the set.	4-6 minutes

Table S2 – Exercises included in the remotely supervised sessions in the web software.

Name	Execution	Training Volume	Progression	Progression Parameter	Approximate Duration
Massage 	Sitting, with leg crossed over the other, massage the sole of your feet with both hands, for 20 seconds. Rub your foot in a circular motion using your thumb. Do the same on the other foot.	1 set of 20 seconds each foot	-	-	40 seconds

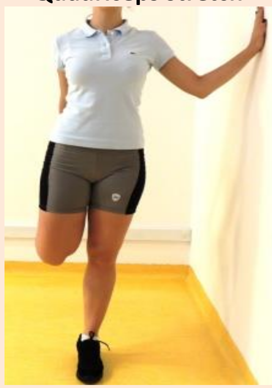
Toes manipulation 	Sitting, with leg crossed over the other, hold each toe and slowly spin side to side, like a screw. Do with all toes.	1 set of 10 times each finger	-	-	1 minute
Feet tapping 	With the heel fixed, tap your foot as fast as possible. Starts seated on a chair, and do with both feet at the same time. After you learn, do the same tapping standing.	1 set of 30 repetitions	1: 1x30 repetitions; 2: 2x30 repetitions; 3: 2x40 repetitions ;	Being able to perform the set without pain or muscle cramp after the completion of the set.	1-2 minutes
Forefoot accend 	Standing, ascend and descend on forefoot. Start standing, using both feet. Use a chair or table to keep balance.	1 set of 30 repetitions	1: 1x30 repetitions; 2: 2x30 repetitions; 3: 2x40 repetitions	Being able to perform the set without pain or muscle cramp after the fulfillment.	1-2 minutes

Invert/Evert symetric 	Sitting, with 90 degrees of knee and ankle flexion, perform symmetrical foot inversion (lifting medial side) and eversion (lifting lateral side).	1 set of 10 repetitions maintaining each position for 1 second.	1: Sitting 1x10 repetitions; 2: Standing 1x10 repetitions 3: Standing 1x20 repetitions	Being able to perform the set without pain or muscle cramp after the completion of the set and without loss of balance.	1-2 minutes
Cotton ball grab  **Rubber ball grab**  **Pen grab** 	While sitting, with the heel in a fixed position, grip the object with the toes, lifting off from the ground and placing it back to its original position. Do the same with the other foot.	1 set of 10 repetitions each foot.	1: 1x10 repetitions with cotton ball; 2:2x10 repetitions with rubber ball; 3: 3x10 repetitions with a pen.	Being able to perform the set without pain or muscle cramp after the completion of the set	3-6 minutes

1-5 toe alternate 	Sitting, with the heel fixed and contacting the floor, alternately pull the hallux and the little toe on the floor. Do it slowly and under complete control.	1 set of 10 repetitions each foot, maintaining finger pressure on the ground for 1 second.	1: Sitting 1x10 repetitions; 2: Standing 1x10 repetitions; 3: Single leg stance 1x10 repetitions.	Being able to perform the set without pain or muscle cramp after the completion of the set and with high control of speed and motion.	2-3 minutes
Toes abduction 	Sitting, with 90 degrees of knee and ankle flexion, abduct and adduct the toes rhythmically.	1 set of 10 repetitions each foot, maintaining 2 seconds abducted and 2 seconds on adducted.	1: Sitting 1x10 repetitions; 2: Standing 2x10 repetitions; 3: Standing 2x10 repetitions maintained for 5 seconds	Being able to perform the set without pain or muscle cramp after the completion of the set and be able to keep the abduction and adduction time.	1-2 minutes
Plantar arch raise 	Sitting, raise the plantar arch in an arch shape. The heel and fingertips should not get off the ground.	1 set of 10 repetitions each foot, maintaining 5 seconds each contraction.	1: Sitting 1x10 repetitions; 2: Standing 1x10 repetitions; 3: Single leg stance 1x10 repetitions.	Being able to perform the set without pain or muscle cramp after the completion of the set.	4-6 minutes

Short-foot exercise 	Sitting, with 90 degrees of knee and ankle flexion, approximate the head of the first metatarsal toward the heel without toe flexion, “shortening” the feet. The forefoot and heel should not get off the ground.	1 set of 10 repetitions each foot, maintaining 5 seconds each contraction.	1: Sitting 1x10 repetitions; 2: Standing 1x10 repetitions; 3: Single leg stance 1x10 repetitions.	Being able to perform the set without pain or muscle cramp after the completion of the set.	4-6 minutes
Toes grasping gait 	Walking “grasping” the toes when they touch the ground. Each step grasp for 3 seconds.	1 set of 10 steps.	1: 1x10 steps; 2: 2x10 steps; 3: 3x10 steps;	Being able to perform the set in the time described and without pain or muscle cramp after the completion of the set.	1-3 minutes
Toes abducted gait 	Walking abducting the toes when the foot touches the ground until take the foot off the ground.	1 set of 3 steps forward and 3 steps backwards.	1: 1x10 repetitions; 2: 2x10 repetitions; 3: 3x10 repetitions;	Being able to perform the set without pain or muscle cramp after the completion of the set.	2-6 minutes

Table S3 – Warm up and stretching exercises**.**

Name	Execution	Training Volume	Approximate Duration
Calf stretch 	Standing in front of a wall, keep one leg in front of the other. The front leg with the knee flexed and the rear leg with the knee extended. Lean forward at the ankle, keeping both heel on the ground, stretching the calf muscles.	1 set of 20 seconds each leg.	40 seconds
Quadriceps stretch 	Standing on one foot, pull the heel towards the bottom, stretching the anterior muscles of the thigh. If necessary, use a wall for support.	1 set of 20 seconds each leg.	40 seconds

Fingertip-to-floor 	Standing, with your back straight, bend your trunk forward, keeping the knee straight, trying to touch the fingertip to the ground.	1 set of 20 seconds each leg.	40 seconds
Lateral stretch (1) 	Standing, with the back straight and with leg crossed over the other, bend the trunk forward, keeping both knees straight, trying to touch the fingertip to the ground.	1 set of 20 seconds each leg.	40 seconds

Adductors stretch 	Sitting, with back straight, knees apart and the sole of feet together, apply gentle pressure to your knees directed to the floor.	1 set of 20 seconds each leg.	40 seconds
Pretzel Stretch 	Lying, with leg crossed over the other, interlace your fingers on the back of the thigh, pulling the leg crossed towards the trunk.	1 set of 20 seconds each leg.	40 seconds
Lateral stretch (2) 	Lying with open arms, flex and adduct the hip directing the knee to the hand of the opposite side.	1 set of 20 seconds each leg.	40 seconds

ANEXO 5 – Additional file 2 do Capítulo V

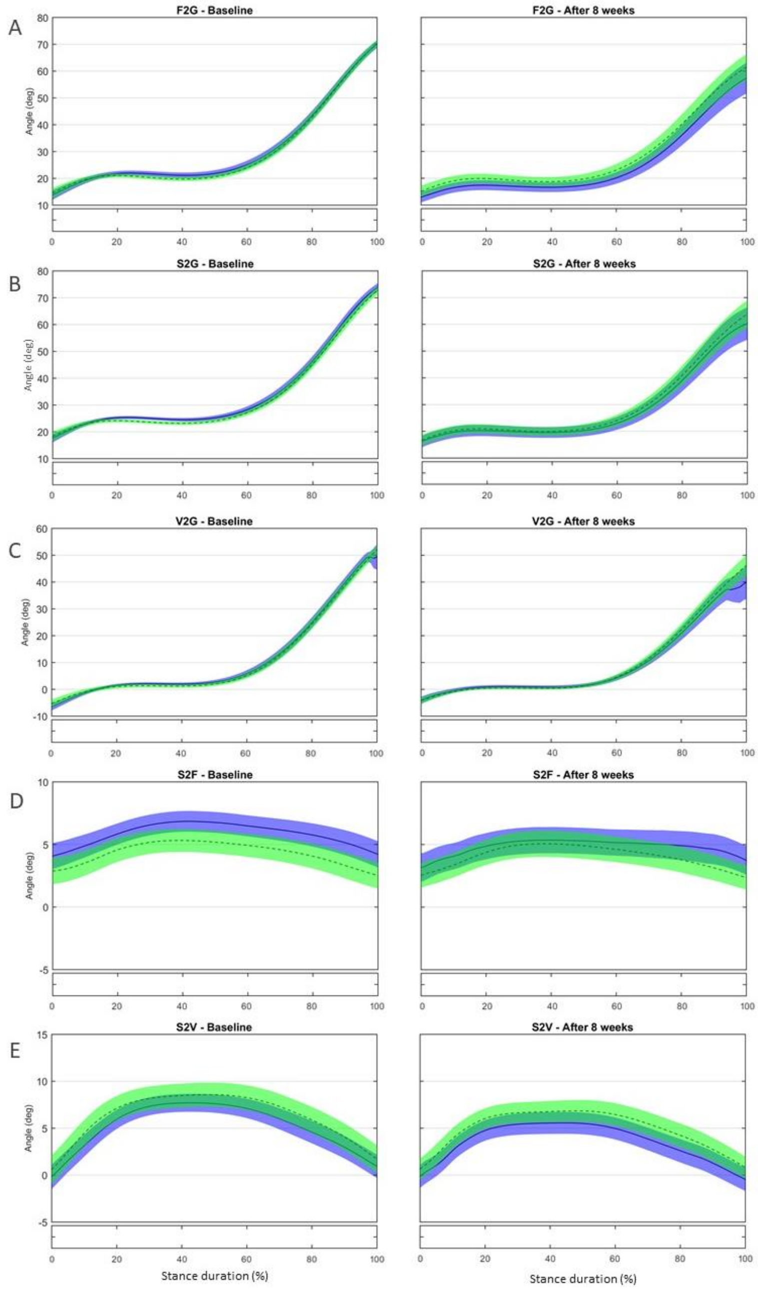


Figure 23 - Mean (±1SD) joint rotation angles during normalized stance phase duration of running in the baseline (left) and after 8 weeks of intervention (right). From top to bottom: Sagittal-plane inclination of the first metatarsal bone to the ground (A), of the second metatarsal bone to the ground (B), and of the fifth metatarsal bone to the ground (C); transverse-plane divergence between first and second metatarsal bones (D) and between fifth and second metatarsal bones (E). Green, CG group; Blue, IG group. The black bar below the graph represents the time during which the differences between the groups occurred ($p<0.05$), what was indicated by the SPM{t} statistic.

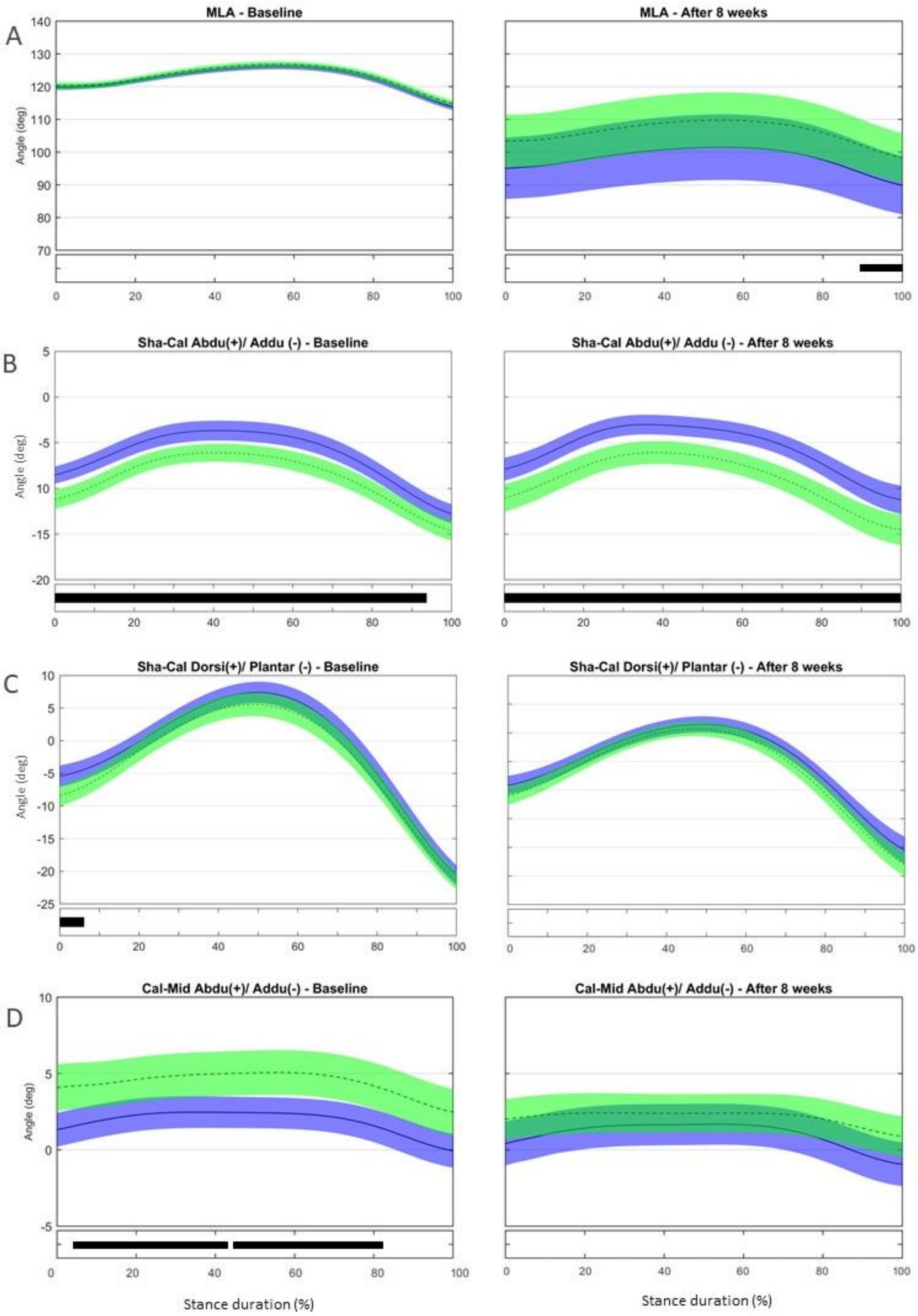


Figure 24 - Mean (±1SD) joint rotation angles during normalized stance phase duration of running in the baseline (left) and after 8 weeks of intervention (right). From top to bottom: Medial longitudinal arch angle (A), transverse-plane rotations between calcaneus and shank (B), sagittal-plane rotations between calcaneus and shank (C), and transverse-plane rotations midfoot and calcaneus (D). Green, CG group; Blue, IG group. The black bar below the graph represents the time during which the differences between the groups occurred (p<0.05), what was indicated by the SPM{t} statistic.

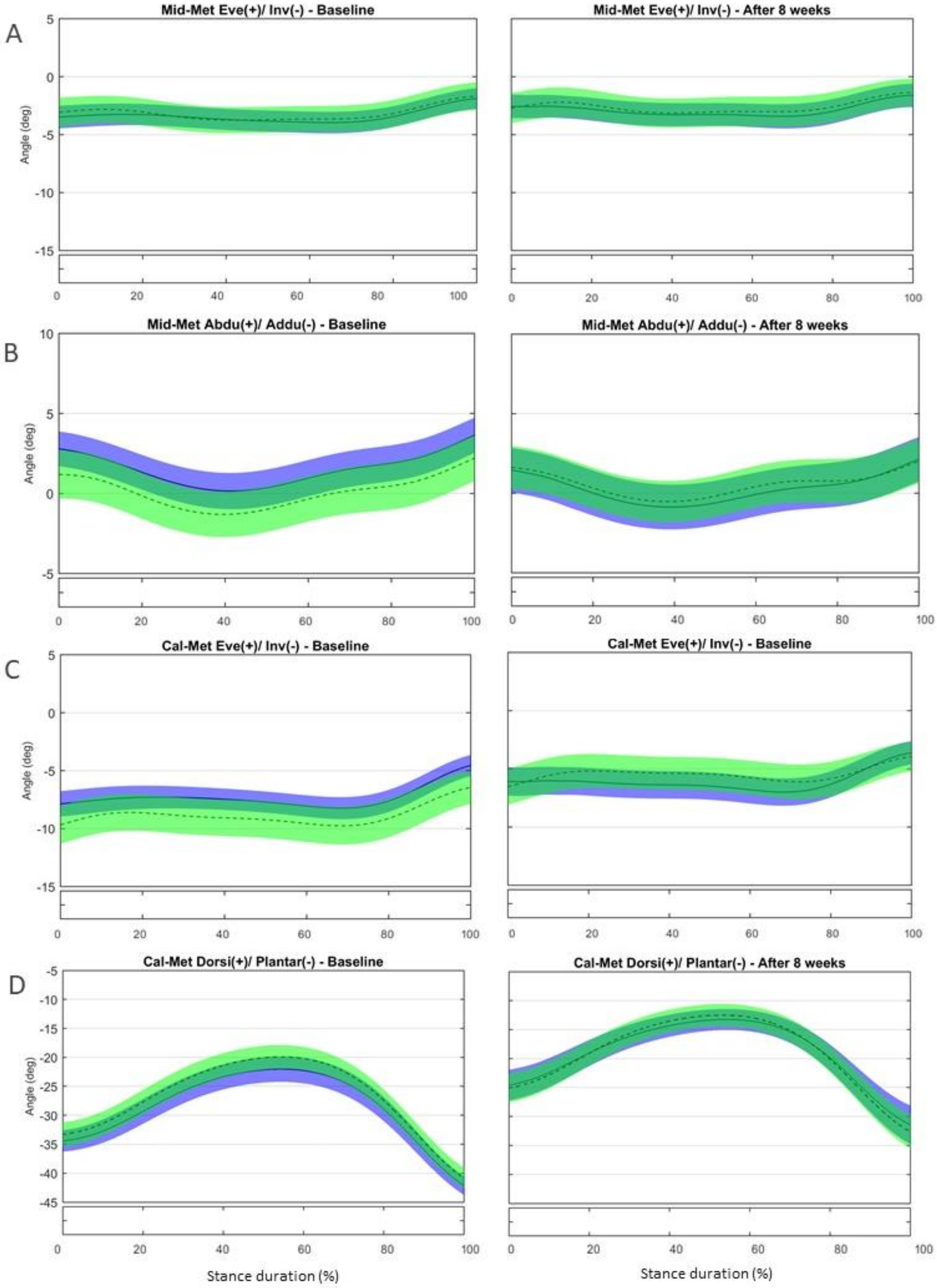


Figure 25 - Mean (±1SD) joint rotation angles during normalized stance phase duration of running in the baseline (left) and after 8 weeks of intervention (right). From top to bottom: frontal-plane rotations between metatarsus and midfoot (A), transverse-plane rotations between metatarsus and midfoot (B), frontal-plane rotations between metatarsus and calcaneus (C), and sagittal-plane rotations between metatarsus and calcaneus (D). Green, CG group; Blue, IG group. The black bar below the graph represents the time during which the differences between the groups occurred (p<0.05), what was indicated by the SPM{t} statistic.

ANEXO 6 - Additional file 3 do Capítulo V

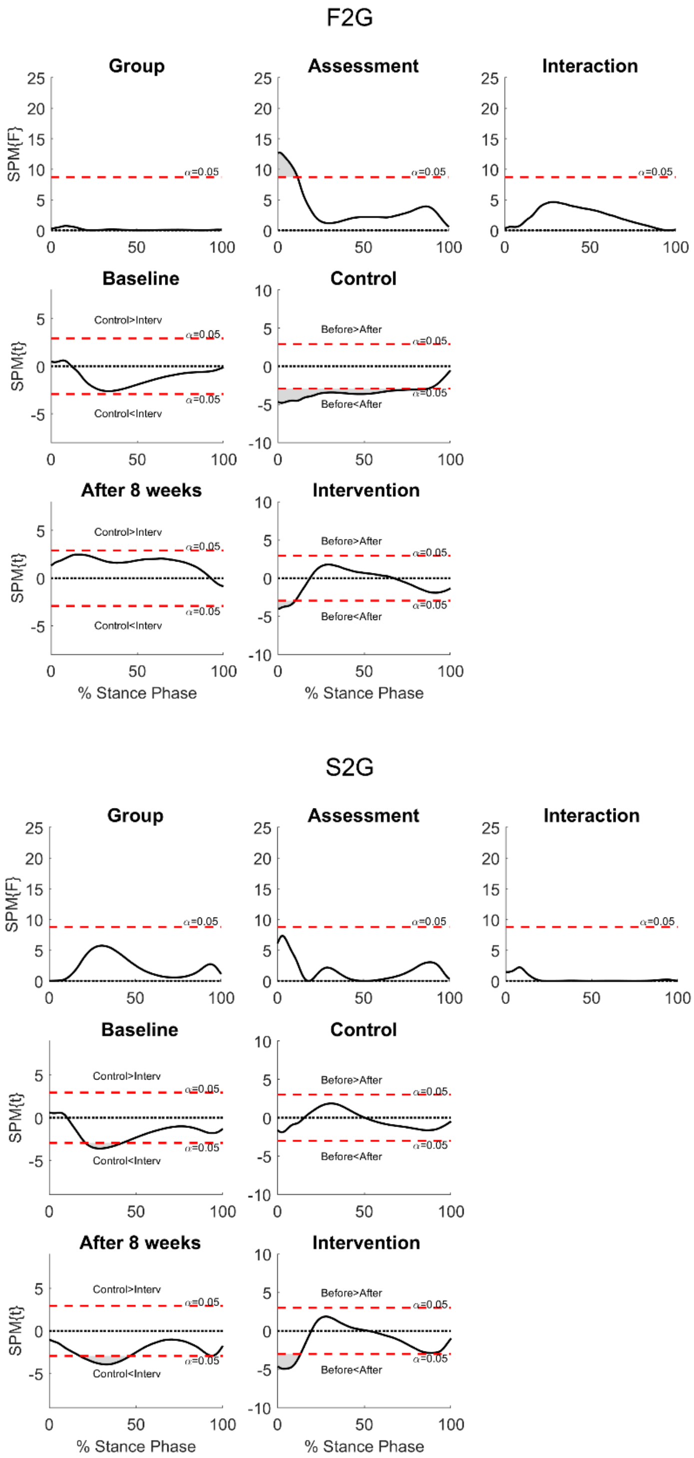


Figure 26 - Statistical parametric mapping results of (top) (sagittal-plane inclination of first metatarsal bone to the ground (F2G) and (bottom) second metatarsal bone to the ground (S2G). Black bars indicate significant differences between both waveforms, where the SPM{t} values exceeded the Sidák corrected alpha level threshold.

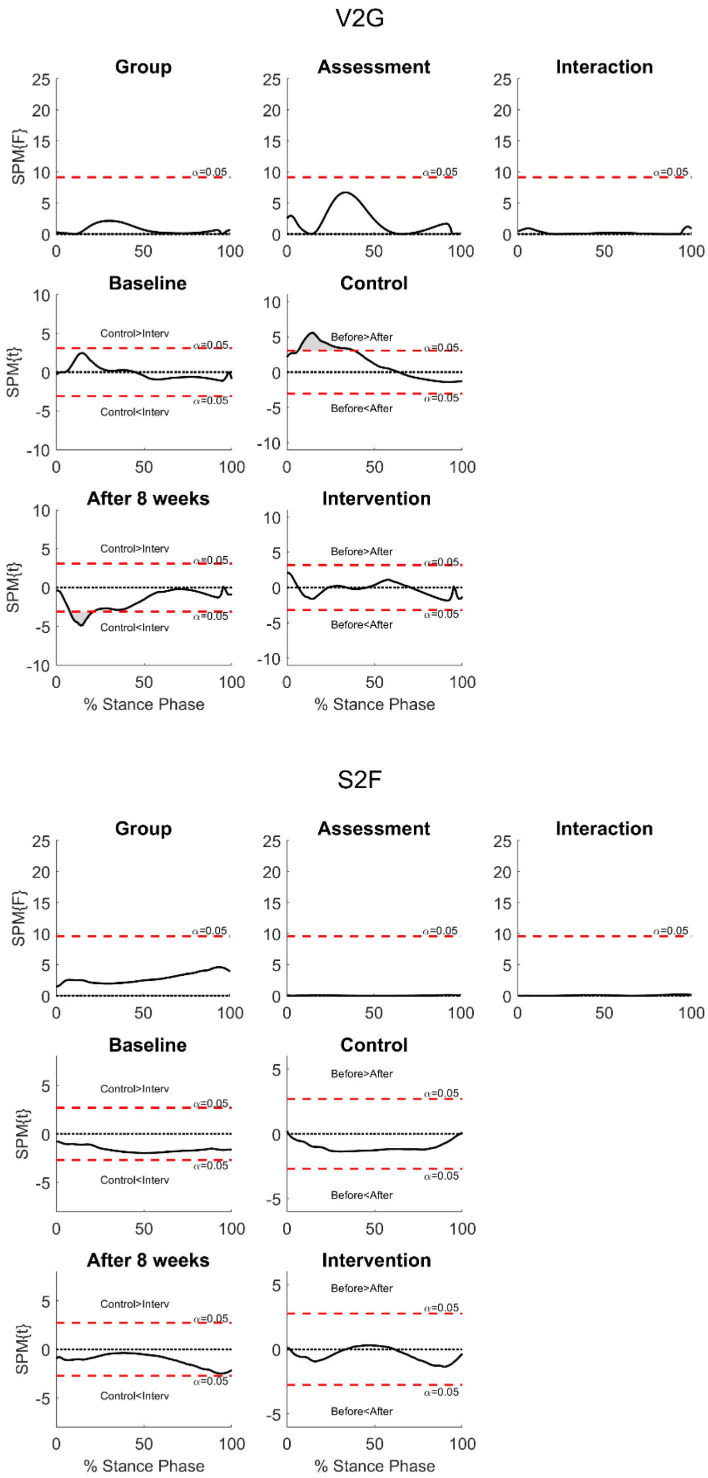


Figure 27 - Statistical parametric mapping results of sagittal-plane inclination of (top) fifth metatarsal bone to the ground (V2G) and transverse-plane divergence between (bottom) first and second metatarsal bones (S2F). Black bars indicate significant differences between both waveforms, where the SPM{t} values exceeded the Sidák corrected alpha level threshold.

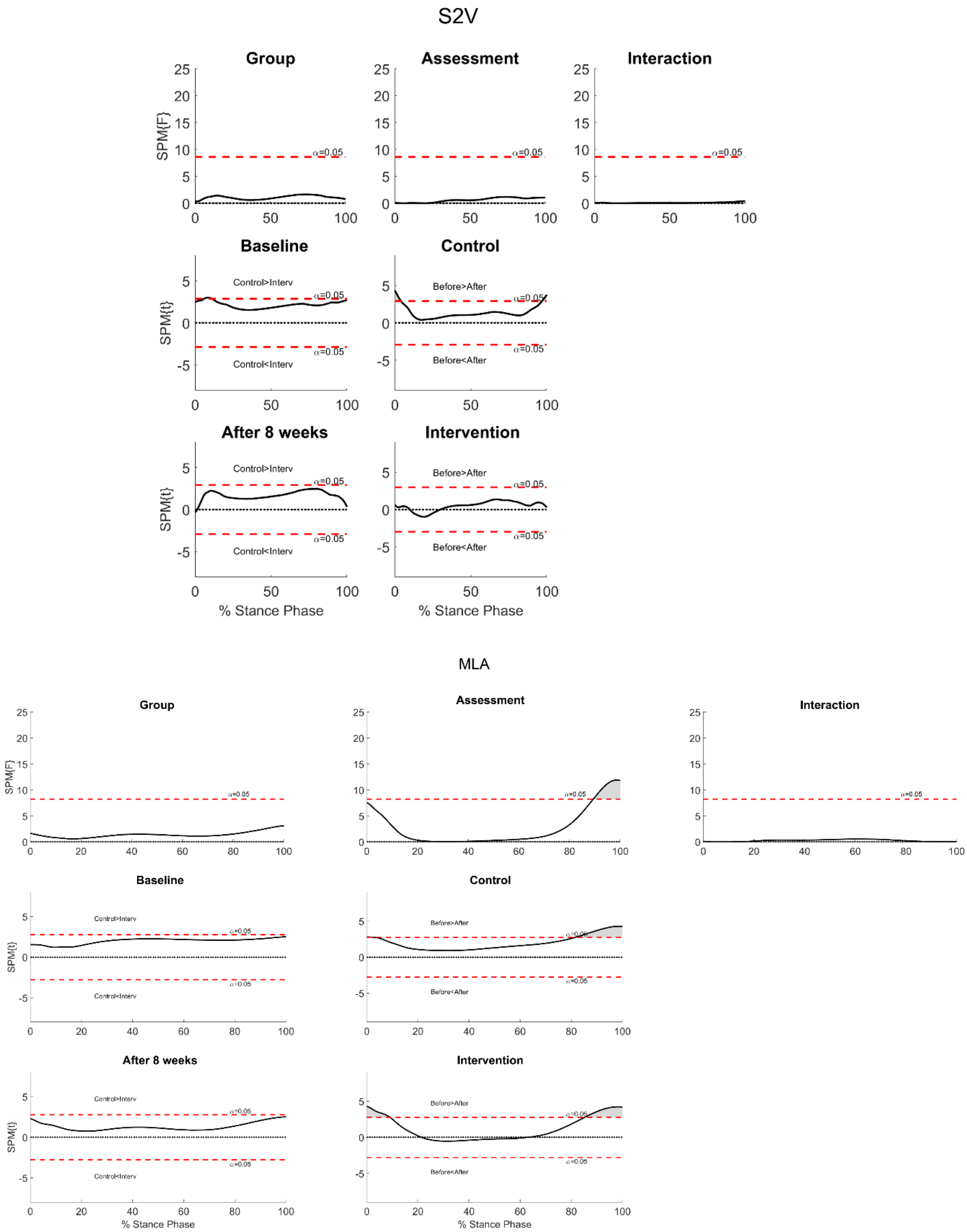


Figure 28 - Statistical parametric mapping results of transverse-plane divergence between (top) second and fifth metatarsal bones (S2V), and (bottom) medial longitudinal arch (MLA). Results were normalized across stance phase (0–100%)Black bars indicate significant differences between both waveforms, where the SPM{t} values exceeded the Sidák corrected alpha level threshold.

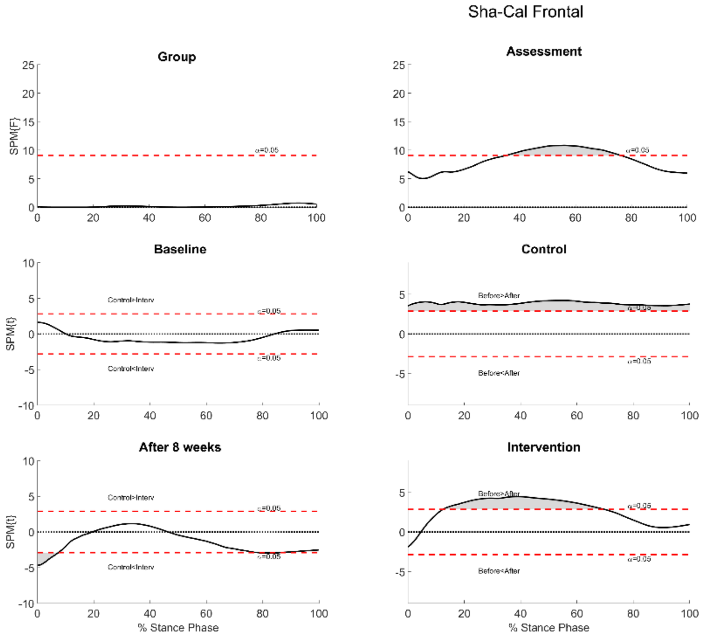


Figure 29 - Statistical parametric mapping results of frontal plane between calcaneus with respect to the shank (Sha-Cal) joint angles time normalized across stance phase (0–100%). Black bars indicate significant differences between both waveforms, where the SPM{t} values exceeded the Sidák corrected alpha level threshold.

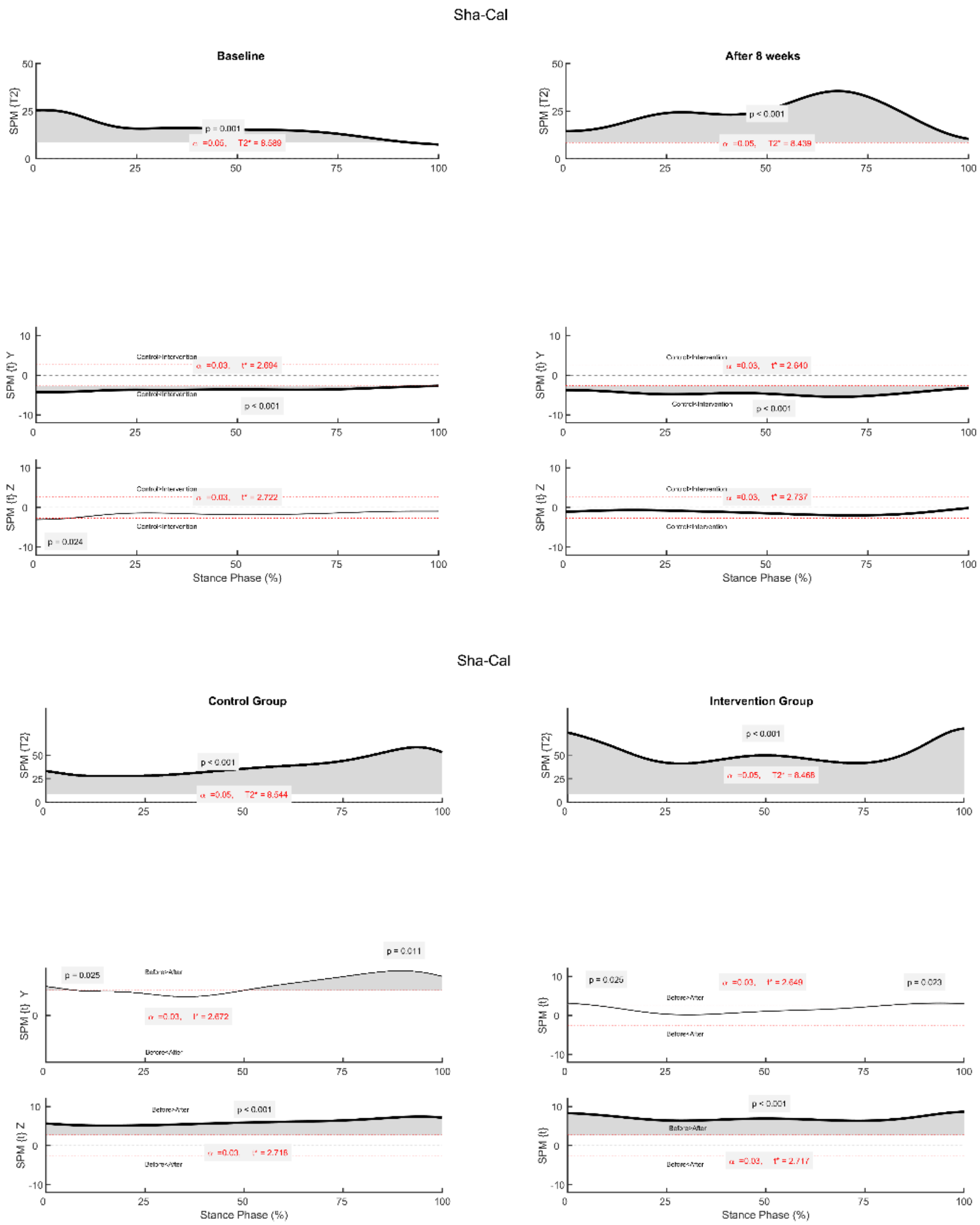


Figure 30 - Statistical parametric mapping (3D vector field SPM analysis) followed by the paired or independent t-test, as a post-hoc tests, results of calcaneus with respect to the shank (Sha-Cal) joint angles time normalized across stance phase (0–100%). Black bars indicate significant differences between both waveforms, where the SPM{t} values exceeded the Sidák corrected alpha level threshold.

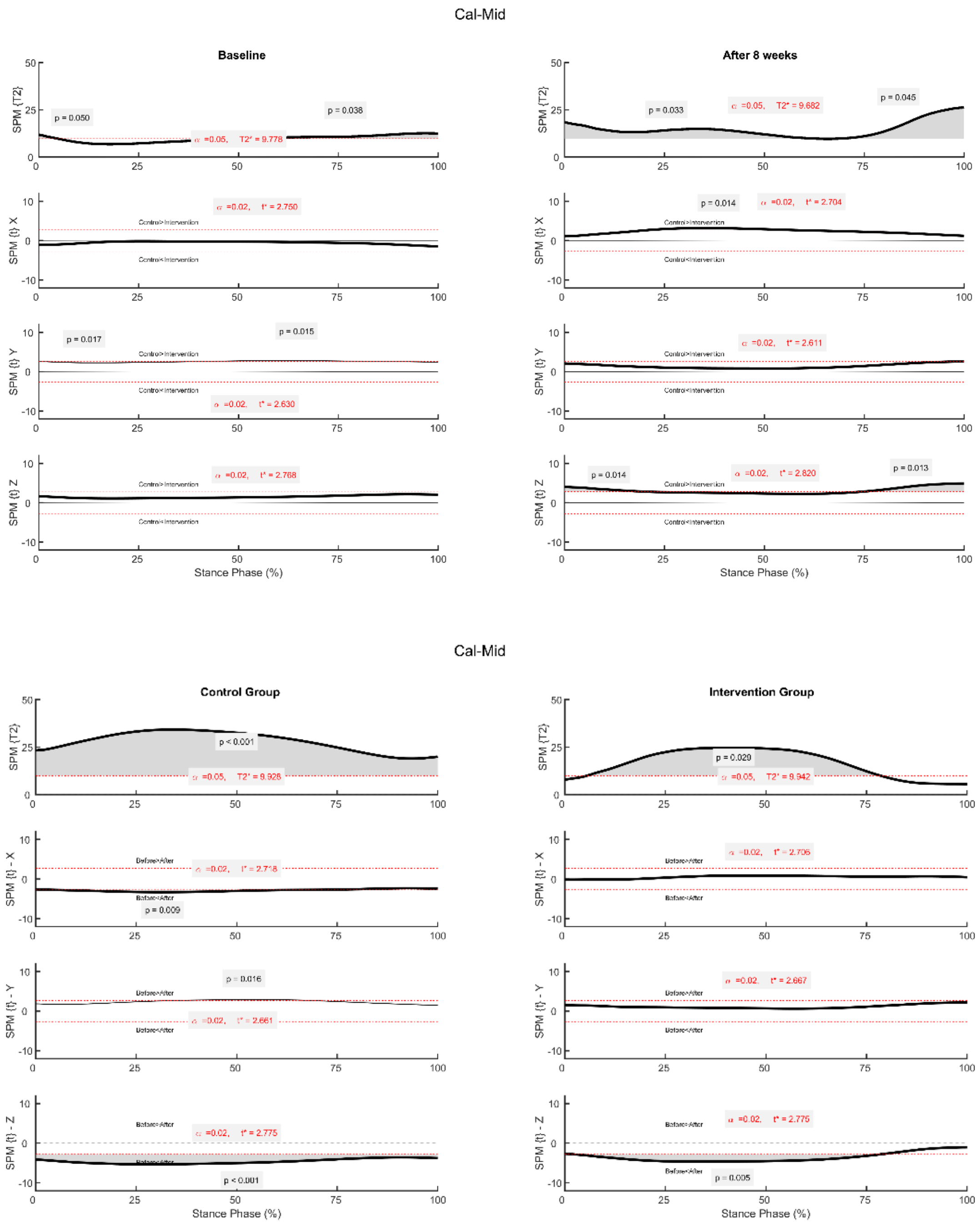


Figure 31 - Statistical parametric mapping (3D vector field SPM analysis) followed by the paired or independent t-test, as a post-hoc tests, results of midfoot with respect to the calcaneus (Cal-Mid) joint angles time normalized across stance phase (0–100%). Black bars indicate significant differences between both waveforms, where the SPM{t} values exceeded the Sidák corrected alpha level threshold.

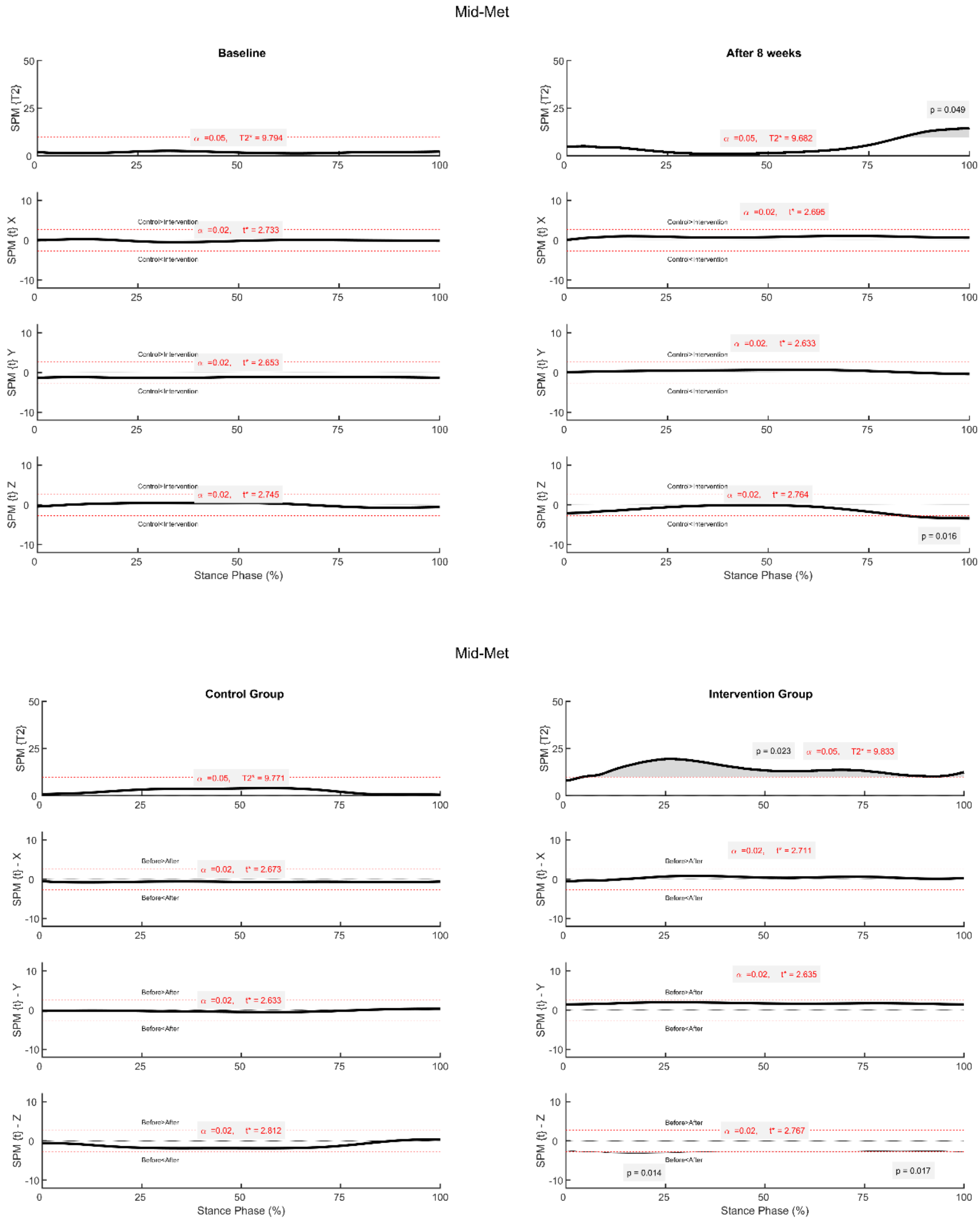


Figure 32 - Statistical parametric mapping (3D vector field SPM analysis) followed by the paired or independent t-test, as a post-hoc tests, results of metatarsus with respect to the midfoot (Mid-Met) joint angles time normalized across stance phase (0–100%). Black bars indicate significant differences between both waveforms, where the SPM{t} values exceeded the Sidák corrected alpha level threshold.

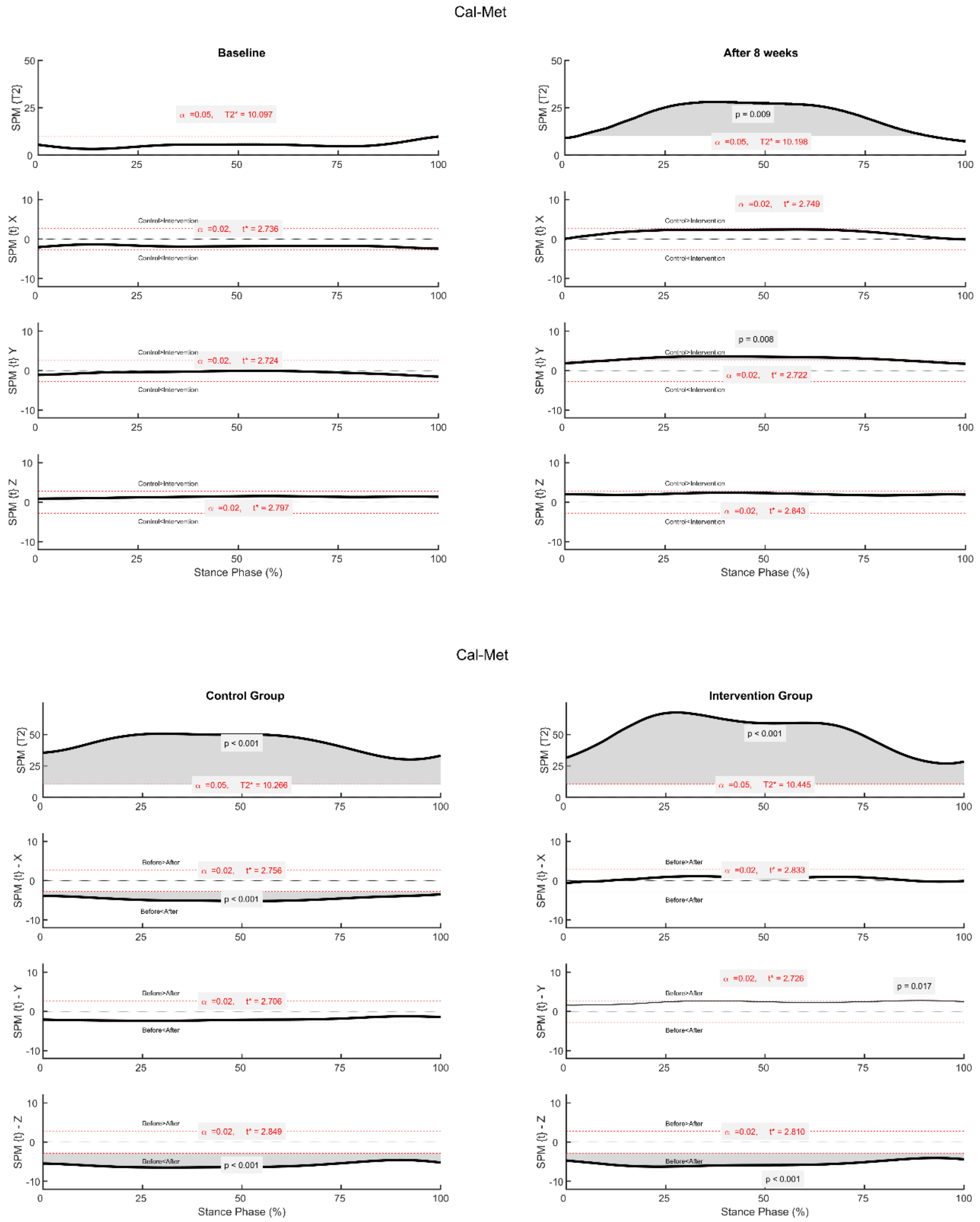
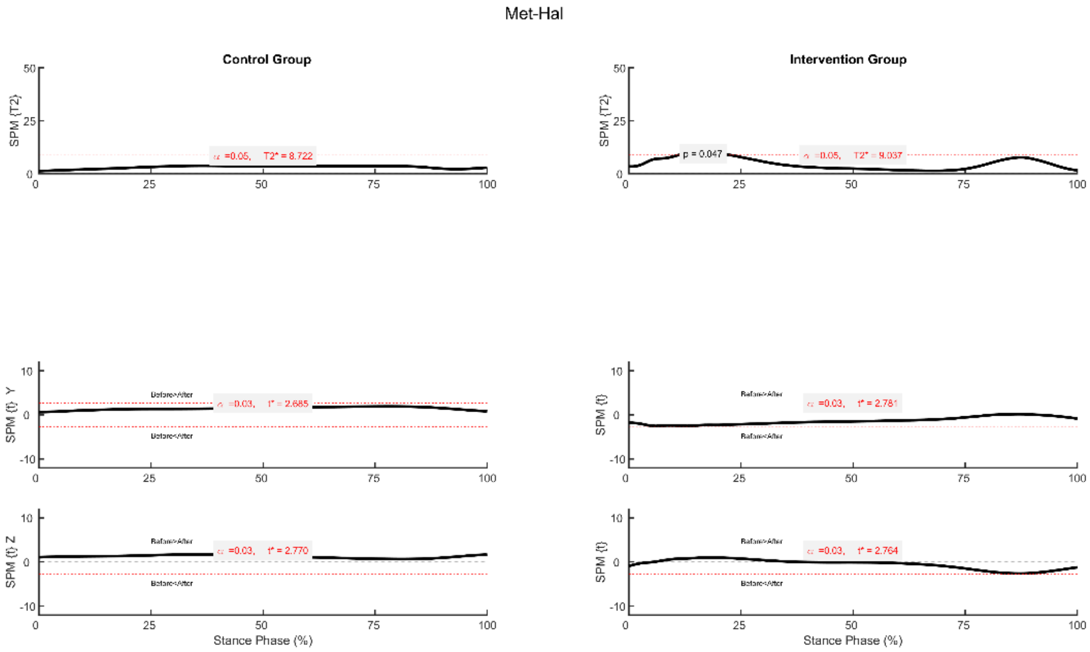


Figure 33 - Statistical parametric mapping (3D vector field SPM analysis) followed by the paired or independent t-test, as a post-hoc tests, results of metatarsus with respect to the calcaneus (Cal-Met) joint angles time normalized across stance phase (0–100%). Black bars indicate significant differences between both waveforms, where the SPM{t} values exceeded the Sidák corrected alpha level threshold.



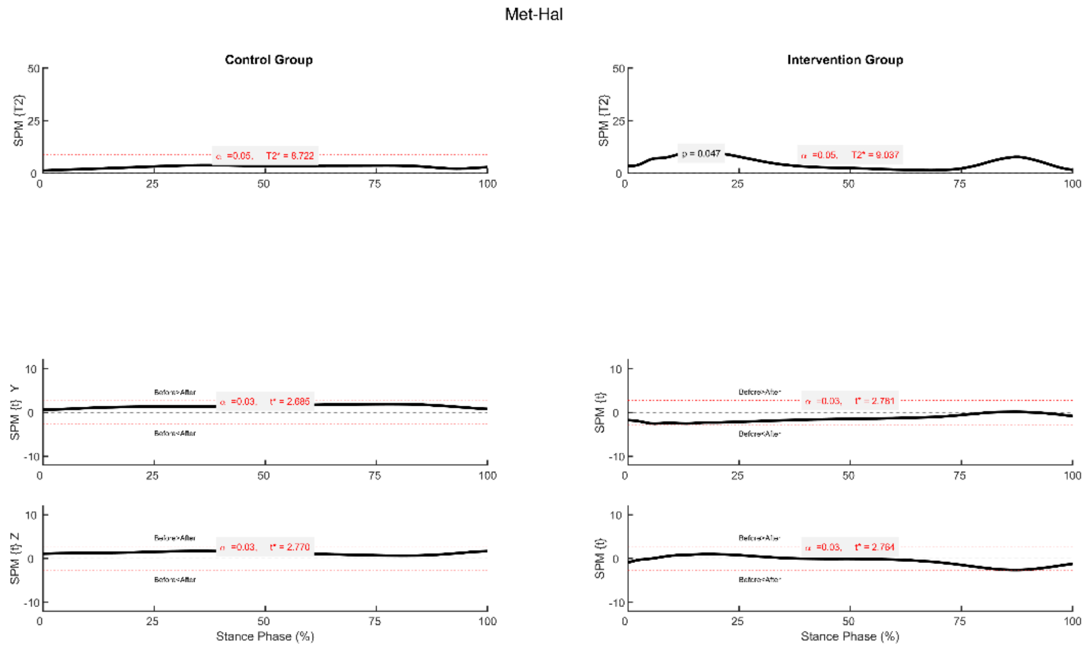


Figure 34 - Statistical parametric mapping (3D vector field SPM analysis) followed by the paired or independent t-test, as a post-hoc tests, results of transverse-plane divergence and sagittal-plane inclination between first metatarsus and hallux angle (Met-Hal). joint angles time normalized across stance phase (0–100%). Black bars indicate significant differences between both waveforms, where the SPM{t} values exceeded the Sidák corrected alpha level threshold.