**UNIVERSIDADE FEDERAL DOS VALES DO JEQUITINHONHA E MUCURI**
**Programa de Pós-Graduação em Reabilitação e Desempenho Funcional**

**Douglas Novaes Bonifácio**

# EFEITO DE UMA PALMILHA COM CUNHA MEDIAL SOBRE A CINEMÁTICA, CINÉTICA E ATIVAÇÃO MUSCULAR DOS MEMBROS INFERIORES E PELVE DE VOLUNTÁRIOS HÍGIDOS DURANTE A TAREFA DE DESCIDA DE DEGRAU.

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Dissertação apresentado ao Programa de Pós-Graduação em Reabilitação e Desempenho Funcional como requisito para a obtenção de título de Mestre.

**Orientador:** Renato Guilherme Trede Filho
**Co-Orientador:** Jim Richards

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DOUGLAS NOVAES BONIFÁCIO

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Dissertação apresentada ao MESTRADO EM REABILITAÇÃO E DESEMPENHO FUNCIONAL, nível de MESTRADO como parte dos requisitos para obtenção do título de MAGISTER SCIENTIAE EM REABILITAÇÃO E DESEMPENHO FUNCIONAL

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DIAMANTINA

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**RESUMO**

Diariamente, o sistema locomotor humano é desafiado a transpor diferentes níveis de terrenos. Subir e descer escadas é uma atividade comum da vida diária, podendo ser um obstáculo quando a função motora está comprometida. Pacientes com dor patelofemoral se queixam de dor retropatelar agravada por atividades como subida e descida de degraus, sendo a dor maior durante a descida de degraus. A necessidade de controle durante a tarefa de descida de degrau e conseguida através da contração muscular excêntrica. Na descida de degrau, os indivíduos devem transportar ativamente o centro de gravidade para frente e resistir a gravidade durante a fase de descida controlada. Estudos vêm investigando respostas das alterações dos padrões fisiológicos de movimentos nos pés, por acreditarem que essas alterações podem originar compensações aos membros inferiores. A pronação excessiva tem sido relacionada a numerosas alterações funcionais nos membros inferiores, indicando lesões de uso excessivos que afetam o quadril, joelho, tornozelo e pé. Os clínicos geralmente concordam que algumas formas de órteses ou calçados ortopédicos especializados podem controlar a pronação excessiva. Diante disso, o presente estudo tem como objetivo avaliar o efeito de uma palminha com cunha medial a cinemática, cinética e a ativação muscular dos membros inferiores e pelve de voluntários hígidos durante a tarefa de descida degrau. O estudo foi dividido em três capítulos: 1) revisão de literatura; 2) artigo científico; 3) considerações finais. A revisão de literatura aborda inicialmente a anatomia de membros inferiores e pelve, seguida das fases e subfase da tarefa de decida de degrau. Em seguida foram abordados a cinemática, cinética e ativação muscular de membros inferiores e pelve durante a tarefa de decida de degrau e uma discussão sobre o uso de palmilhas com arco longitudinal medial e sua resposta de forma ascendente aos membros inferiores. O artigo científico avaliou a influência e os benefícios das órteses plantares quanto a cinemática, cinética e ativação muscular na tarefa de decida de degrau. Como consideração final apontou que as órteses plantares podem alterar a mecânica articular do pé e dos membros inferiores, fornecendo benefícios funcionais durante a tarefa de descida de degrau.

**Palavras-chave:** palmilha, biomecânica, cinética, cinemática, degrau.

# ABSTRACT

Daily, the human locomotor system is challenged to cross different levels of terrain. Going up and down stairs is a common activity of daily living and can be an obstacle when motor function is compromised. Patients with patellofemoral pain complain of retropatellar pain aggravated by activities such as ascent and descent of stairs, with pain being greater during the descent of stairs. The need for control during the step descent task is achieved through eccentric muscle contraction. In the descent of the step, the individuals must actively transport the center of gravity forward and resist gravity during the phase of controlled descent. Studies have been investigating responses of changes in physiological patterns of movement in the feet, believing that these changes can lead to compensations to the lower limbs. Excessive pronation has been linked to numerous functional changes in the lower limbs, indicating excessive use injuries affecting the hip, knee, ankle and foot. Clinicians generally agree that some forms of specialized orthopedic orthoses or footwear can control excessive pronation. Therefore, the present study aims to evaluate the effect of a limb wedge with the kinematics, kinetics and muscular activation of the lower limbs and pelvis of healthy volunteers during the task of descending step. The study was divided into three chapters: 1) literature review; 2) scientific article; 3) final considerations. The literature review initially addresses the anatomy of the lower limbs and pelvis, followed by the phases and sub-phase of the step deciding task. Kinematics, kinetics and muscular activation of the lower limbs and pelvis were discussed during the decision-making task and a discussion about the use of insoles with a medial longitudinal arch and its response ascending to the lower limbs. The scientific article evaluated the influence and benefits of plantar orthoses regarding kinematics, kinetics and muscular activation in the task of deciding step. As a final consideration it was pointed out that the plantar orthoses can alter the joint mechanics of the foot and lower limbs, providing functional benefits during the step descent task.



**Words key:** Insole, biomechanics, kinetic, kinematic, step.

**SUMÁRIO**

# CAPÍTULO 1 – Revisão de literatura

## Introdução

Diariamente, o sistema locomotor humano é desafiado a transpor diferentes níveis de terrenos (COWIE; BRADDICK; ATKINSON, 2008). Subir e descer escadas é uma atividade comum da vida diária (PROTOPAPADAKI *et al.*, 2007), podendo ser um obstáculo quando a função motora está comprometida (ALDRIDGE WHITEHEAD; RUSSELL ESPOSITO; WILKEN, 2016; CLUFF; ROBERTSON, 2011; KOWALK; DUNCAN; VAUGHAN, 1996). Segundo estudo de Reis *et al* (2009), a dor patelofemoral pode ser provocada ou acentuada durante atividades de vida diária, como subir e descer escadas, agachar e andar de bicicleta, devido ao aumento da sobrecarga articular durante movimentos de flexão e extensão do joelho (REIS *et al.*, 2009). Da mesma forma, estudos apontam que pacientes com dor patelofemoral se queixam de dor retropatelar agravada por atividades como subida e descida de degraus (BRECHTER; POWERS, 2002; NAKAGAWA *et al.*, 2011). Segundo Brindle *et al* (2003), a dor tende a ser maior durante a descida de degraus (BRINDLE; MATTACOLA; MCCRORY, 2003). Selfe *et al* (2008) e Alcock *et al* (2015), apontam que a descida de degraus é mais desafiadora do que a subida de degraus devido ao nível de controle muscular excêntrico necessário (ALCOCK; O’BRIEN; VANICEK, 2015; SELFE *et al.*, 2008). Antes de discutirmos sobre a biomecânica da tarefa de descida de degraus, é importante uma compreensão da anatomia das extremidades inferiores.

O complexo tornozelo-pé é uma elaborada estrutura cinemática composta por 26 ossos, interligados com uma grande quantidade de ligamentos e músculos, atravessados por tendões musculares longos e fáscias (CHAN; RUDINS, 1994; DUERINCK *et al.*, 2014). Suas funções são: absorver parte da força de reação do solo e impulsionar o corpo em diferentes direções durante a marcha (FIOLKOWSKI *et al.*, 2003; SILVA, V. R., 2015). Além disso, os pés necessitam ser suficientemente maleáveis após o contato inicial da marcha com o objetivo de dissipar energia e de se adaptar às irregularidades do solo (FIOLKOWSKI *et al.*, 2003; SILVA, V. R., 2015). Posteriormente, este segmento corporal deve ser relativamente rígido para ao final da fase de apoio da marcha agir como uma alavanca, impulsionando o corpo para frente (FIOLKOWSKI *et al.*, 2003; SILVA, V. R., 2015).

A articulação do tornozelo se refere principalmente a articulação talocrural que é formada pela articulação da face superior da tróclea do tálus com a cavidade retangular formada pela extremidade distal da tíbia e ambos os maléolos, mas também inclui duas articulações relacionadas: a articulação tibiofibular que é formada pela cabeça da fíbula e pela face póstero-lateral do côndilo lateral da tíbia e a sindesmose tibiofibular que é formada pela articulação da face medial convexa da parte distal da fíbula, com a incisura fibular côncava da tíbia (HOUGHTON, 2008). O segmento pé se refere a todas as estruturas distais à tíbia e fíbula. O pé pode ser dividido em três regiões, cada uma consistindo em um conjunto de ossos e de uma ou mais articulações (HOUGHTON, 2008). O retropé consiste do tálus, da articulação subtalar e do calcâneo (FRASER; FEGER; HERTEL, 2016). A articulação subtalar é uma articulação entre a face inferior do tálus e a face superior do calcâneo e é composta pelas articulações talocalcaneonavicular e talocalcânea (CHAN; RUDINS, 1994; FRASER; FEGER; HERTEL, 2016). O mediopé consiste no restante dos ossos tarsais, incluindo a articulação transversa do tarso que é formada pelas seguintes articulações: a articulação talocalcaneonavicular e a articulação calcaneocubóidea, e as articulações intertarsais distais menores (FRASER; FEGER; HERTEL, 2016; HOUGHTON, 2008). O antepé consiste nos ossos metatarsais e falanges, incluindo todas as articulações distais às articulações tarsometatarsais (FRASER; FEGER; HERTEL, 2016; HOUGHTON, 2008).

A terminologia usada para descrever os movimentos do tornozelo e do pé incorpora dois conjuntos de definições: um conjunto fundamental e um conjunto aplicado (KENDALL, FP. MCCREARY, EK. PROVANCE, PG, RODGERS, MM, ROMANI, 2007). A terminologia fundamental descreve o movimento do pé e do tornozelo que ocorre em ângulos retos com três eixos padrões de rotações (KENDALL, FP. MCCREARY, EK. PROVANCE, PG, RODGERS, MM, ROMANI, 2007). A dorsiflexão e a flexão plantar descrevem o movimento que é paralelo ao plano sagital, em torno de um eixo medio-lateral de rotação. Eversão e inversão descrevem o movimento paralelo ao plano coronal em torno de um eixo ântero-posterior de rotação. Abdução e adução descrevem o movimento no plano horizontal (transverso) em torno de um eixo vertical (supero-inferior) de rotação (DUGAN; BHAT, 2005; KENDALL, FP. MCCREARY, EK. PROVANCE, PG, RODGERS, MM, ROMANI, 2007). A terminologia aplicada é usada para descrever os movimentos que ocorrem perpendicular a um eixo oblíquo de rotação (KENDALL, FP. MCCREARY, EK. PROVANCE, PG, RODGERS, MM, ROMANI, 2007). Pronação descreve um movimento que possui elementos de eversão do calcâneo, abdução e dorsiflexão do tálus. Já a supinação descreve um movimento que possui elementos de inversão do calcâneo, adução e flexão plantar do tálus (DUGAN; BHAT, 2005; HOUGHTON, 2008; KENDALL, FP. MCCREARY, EK. PROVANCE, PG, RODGERS, MM, ROMANI, 2007; SCHUSTER; CHRIS COETZEE; STOVITZ, 2009).

Os eixos mecânicos do tornozelo e do pé não são perpendiculares a nenhum dos planos cardinais, todos os movimentos são essencialmente triplanares (CHAN; RUDINS, 1994). A articulação talocrural é uniaxial e descrita como um dorsiflexor e flexor plantar puro, o seu eixo passa através do maléolo medial, cabeça do tálus e o maléolo lateral (CHAN; RUDINS, 1994; FRASER; FEGER; HERTEL, 2016). Na articulação subtalar ocorrem a maior parte dos movimentos de eversão e inversão (CHAN; RUDINS, 1994). Por seu eixo ser análogo a uma dobradiça oblíqua, os movimentos de inversão e eversão são transmitidos ao aspecto superior do tálus e provocam rotações mediais ou laterais da perna (CHAN; RUDINS, 1994).

A articulação transversa do tarso é uma junção entre as articulações talonavicular e calcaneocubóide, e os movimentos entre o antepé e o retropé ocorrem nesta articulação (CHAN; RUDINS, 1994; FRASER; FEGER; HERTEL, 2016). Na articulação transversa do tarso, um par de eixos funcionais são descritos, variando de acordo com a tarefa e com base na congruência da articulação calcaneocubóide (CHAN; RUDINS, 1994; FRASER; FEGER; HERTEL, 2016). O eixo longitudinal é paralelo ao eixo da articulação subtalar e fornece eversão-abdução e inversão-adução, e permite que o antepé gire o oposto do mediopé no plano transversal (CHAN; RUDINS, 1994; FRASER; FEGER; HERTEL, 2016). O eixo oblíquo é próximo ao eixo da articulação do tornozelo e fornece a dorsiflexão e flexão plantar (CHAN; RUDINS, 1994). Segundo Fraser *et al* (2016), o movimento artrocinemático sobre o eixo oblíquo, produz deformações nos arcos longitudinais plantares (FRASER; FEGER; HERTEL, 2016).

Os ossos do pé são unidos por ligamentos de modo que, no pé típico, nenhum dos ossos entre o calcâneo e as cabeças dos metatarsos, transmitem o peso diretamente ao chão (CHAN; RUDINS, 1994). Assim, a descarga de peso no tálus é transmitida ao calcâneo no retropé e às cabeças dos metatarsos no antepé (CHAN; RUDINS, 1994). O formato dos ossos, combinada com suporte ligamentoso e, em menor grau, muscular, formam dois arcos longitudinais, medial e lateral, e um arco transversal (CHAN; RUDINS, 1994). Os arcos protegem o pé redistribuindo a pressão e possibilitam um pé agindo de forma rígida ou móvel em diferentes fases do movimento (CHAN; RUDINS, 1994).

O arco longitudinal medial é a estrutura principal de sustentação de peso e de absorção de choque no pé (FIOLKOWSKI *et al.*, 2003). O arco longitudinal medial consiste nos ossos calcâneo, tálus e navicular, três cuneiformes e o primeiro, segundo e terceiro ossos metatarsais e é sustentado por duas forças principais: suporte passivo que consiste na fáscia plantar, ligamentos plantares longos e curtos e estabilidade das articulações tarsometatarsais, e suporte ativo formado pelos músculos tibial anterior, tibial posterior, flexor longo e curto do hállux e abdutor do hálux (KENDALL, FP. MCCREARY, EK. PROVANCE, PG, RODGERS, MM, ROMANI, 2007; LEE, J. H. *et al.*, 2016). Na posição ereta relaxada, as forças passivas geralmente são suficientes para suportar o arco, porém, durante ações mais dinâmicas como caminhar, correr, descer e subir degraus, são necessárias forças ativas para manutenção do arco longitudinal medial (KENDALL, FP. MCCREARY, EK. PROVANCE, PG, RODGERS, MM, ROMANI, 2007). Segundo Lee *et al* (2016), músculos extrínsecos e intrínsecos do pé suportam o arco longitudinal medial como um estabilizador ativo durante a marcha e caminhada (LEE, J. H. *et al.*, 2016). Em particular, os músculos intrínsecos do pé como abdutor do hálux, flexor curto do hálux e músculos interósseos ajudam a estabilizar o arco do pé durante a propulsão (LEE, J. H. *et al.*, 2016). Outros estudos abordam a ativação de músculos extrínsecos, como os músculos tibial anterior, tibial posterior e fibular longo no controle do arco longitudinal medial (FIOLKOWSKI *et al.*, 2003; MULLIGAN; COOK, 2013; VAN BOERUM; SANGEORZAN, 2003).

O abdutor do hálux é o músculo mais medial dentro da primeira camada de músculos intrínsecos na superfície plantar do pé (JUNG; KOH; KWON, 2011). O músculo abdutor do hálux surge do calcâneo posteromedial e insere no sesamóide interno ou falange proximal, desempenhando um papel importante no suporte ao arco longitudinal medial e no controle da pronação (FIOLKOWSKI *et al.*, 2003; HEADLEE *et al.*, 2008). Em um estudo de Wong (2007), em cadáveres, aponta que o curso anatômico do abdutor do hálux contribui para suas funções de abdutor do hálux, flexor e supinador do primeiro metatarso, inversor do calcâneo e rotador externo da tíbia em conjunto com a elevação do arco longitudinal medial (WONG, 2007). Segundo estudo de Headlee *et al* (2008), a disfunção de qualquer músculo intrínseco que suporta o arco longitudinal medial, pode predispor os indivíduos a lesões de uso excessivo relacionado à pronação excessiva (HEADLEE *et al.*, 2008). Segundo o mesmo estudo, a queda do navicular aumenta significativamente com a fadiga do músculo abdutor do hálux e com a pronação excessiva dos pés (HEADLEE *et al.*, 2008). Estudos usam a queda do navicular como uma medida clínica para estimar a pronação (HOFFMAN *et al.*, 2015; MATTACOLA; EDITOR; KELLY, 2003). A queda do navicular é definida como a distância entre a altura do navicular com a articulação subtalar no neutro durante o suporte de peso em uma posição relaxada (MATTACOLA; EDITOR; KELLY, 2003). A articulação subtalar está em neutra quando o pé não está supinado ou pronado (MATTACOLA; EDITOR; KELLY, 2003).

O músculo tibial anterior está entre os músculos extrínsecos no controle do arco longitudinal medial, e é conhecido como extensor dorsal do pé e do tornozelo (WILLEGGER *et al.*, 2017). O músculo tibial anterior se insere na face medial do primeiro cuneiforme e a base do primeiro metatarso (FENNELL; PHILLIPS, 1994). O tibial anterior tem função no controle excêntrico do contato de calcanhar, pronto para a aceitação de peso da marcha, além de influenciar no controle da velocidade do movimento de eversão do retropé no plano coronal (HUNT; SMITH; TORODE, 2001b). Segundo Hunt *et al* (2001), o movimento do antepé também será influenciado pelo tibial anterior, pois, suas ações são descritas como elevação e rotação lateral do primeiro metatarso e cuneiforme medial, elevando o arco longitudinal medial durante o movimento de propulsão da marcha (HUNT; SMITH; TORODE, 2001a). Um estudo de Murley *et al* (2009), investigando se a postura do pé influencia na atividade eletromiográfica durante a marcha, aponta que durante a fase de apoio da marcha, o músculo tibial anterior apresentou maior amplitude eletromiográfica no grupo de indivíduos com arco plano em relação ao grupo de indivíduos com arco normal (MURLEY; MENZ; LANDORF, 2009).

O antepé inclui todas as articulações distais às articulações tarsometatarsais, e essas articulações são divididas funcionalmente em cinco raios (CHAN; RUDINS, 1994; HOUGHTON, 2008). O primeiro raio consiste do primeiro metatarso com o cuneiforme medial e realiza movimentos combinados de inversão com dorsiflexão e eversão com flexão plantar, com contribuição mínima para abdução e adução (GLASOE; YACK; SALTZMAN, 1999). Segundo estudo de Cornwall *et al* (2006), o primeiro raio desempenha um papel fundamental ao fornecimento de estabilidade e manutenção da integridade estrutural do pé durante atividades de suporte de peso (CORNWALL *et al.*, 2006). Além disso, destaca-se a importância funcional do primeiro raio com o músculo tibial anterior para manter a estabilidade do arco longitudinal medial durante a fase de impulso do ciclo da marcha (CORNWALL *et al.*, 2006). Segundo estudos de Glasoe *et* *al* (1999) e Cornwall *et al* (2006), a hipomobilidade do primeiro raio pode comprometer a função do arco longitudinal medial devido às altas pressões plantares impostas abaixo da primeira cabeça metatarsiana, limitando o movimento do pé durante a caminhada (CORNWALL *et al.*, 2006; GLASOE; YACK; SALTZMAN, 1999). Já a hipermobilidade do primeiro raio pode apresentar alterações biomecânicas diferentes, devido ao movimento dorsal excessivo e consequente redução ao suporte do arco longitudinal medial (CORNWALL *et al.*, 2006). Essa redução de suporte diminui a capacidade do pé para impulsionar efetivamente o corpo para frente durante a marcha, além de induzir a pronação do pé, sobrecarregando o segundo raio (ALLEN *et al.*, 2004; CORNWALL *et al.*, 2006). O segundo ao quarto raio permitem movimentos de dorsiflexão e flexão plantar, e eles consistem do segundo metatarso com o cuneiforme intermédio, terceiro metatarso com o cuneiforme lateral, e quarto metatarsos (CHAN; RUDINS, 1994). O quinto raio resulta na pronação e supinação entre o quinto metatarso e o cuboide (CHAN; RUDINS, 1994).

O Pé plano é uma condição crônica que inclui um achatamento do arco longitudinal medial, retropé valgo e abdução do mediopé sobre o retropé, sendo associados com a pronação excessiva do pé durante a marcha (JUNG; KOH; KWON, 2011), A pronação excessiva possui múltiplas causas, como o formato anormal ou mobilidade alterada dos ossos tarsais, a rotação medial excessiva do fêmur, fraqueza muscular generalizada e/ou flexibilidade reduzida e a frouxidão nas estruturas que normalmente suportam e controlam o arco longitudinal medial (EUSTACE *et al.*, 1994; KENDALL, FP. MCCREARY, EK. PROVANCE, PG, RODGERS, MM, ROMANI, 2007; LEE, J.; YOON; CYNN, 2016; NGUYEN *et al.*, 2010). O mecanismo da pronação ocorre imediatamente após a fase de contato do calcanhar, onde a articulação talocrural dorsifletida e a articulação talocalcânea ligeiramente supinada rapidamente realizam a flexão plantar e a pronação, respectivamente (KENDALL, FP. MCCREARY, EK. PROVANCE, PG, RODGERS, MM, ROMANI, 2007). O calcâneo em eversão, como resultado da força de reação do solo, empurra simultaneamente a cabeça do tálus, medialmente, no plano horizontal e, inferiormente, no plano sagital (LUDWIG; KELM; FRÖHLICH, 2016). Este movimento do tálus em relação ao calcâneo, abduz e realiza a dorsiflexão da articulação talocalcânea, sendo movimentos consistentes com a pronação (KENDALL, FP. MCCREARY, EK. PROVANCE, PG, RODGERS, MM, ROMANI, 2007; LUDWIG; KELM; FRÖHLICH, 2016). Durante a fase de contato inicial, a tíbia e a fíbula, e em menor extensão o fêmur, giram internamente após o contato inicial do calcanhar, isso devido à configuração da articulação talocrural, onde a rotação medial da parte inferior da perna conduz a articulação talocalcânea a pronação (DUGAN; BHAT, 2005).

Estudos vêm investigando respostas das alterações dos padrões fisiológicos de movimentos nos pés, pois acreditam que essas alterações podem originar compensações aos membros inferiores durante atividades funcionais, podendo ter associações entre os fatores de risco e a ocorrência de lesões (GUIMARÃES *et al.*, 2006; JAMES; BATES; OSTERNIG, 1978; VERDEJO; MILLS, 2002). Segundo estudo de Davis *et al* (2005), o mau alinhamento nas estruturas do antepé, mediopé e retropé, podem levar a movimentos compensatórios ascendentes nas extremidades inferiores e a lesões (DAVIS, 2005). Chuter *et al* (2012), em uma revisão de literatura abordando as contribuições proximais e distais para lesões nas extremidades inferiores, aponta que movimentos como a pronação do pé excessiva ou prolongada têm sido relacionada a numerosas alterações funcionais nos membros inferiores, indicando lesões de uso excessivos que afetam o quadril, joelho, tornozelo e pé (CHUTER; JANSE DE JONGE, 2012). Movimentos como de supinação e pronação excessiva do pé podem ser transferidos de forma ascendente aos membros inferiores, sobretudo devido ao ângulo oblíquo da articulação subtalar, podendo gerar aumento da demanda sobre estruturas como o ligamento cruzado anterior e a articulação patelofemoral (BITTENCOURT, 2010; BOLDT *et al.*, 2013; SOUZA, THALES R. *et al.*, 2014). Segundo estudo de Riskowski *et al* (2013), dores nas articulações das extremidades inferiores são altamente prevalentes (RISKOWSKI *et al.*, 2013). As estimativas sugerem que até 40% das mulheres e 30% dos homens sofrem de dores nas extremidades inferiores, sendo maior em adultos mais velhos, mulheres e indivíduos obesos (RISKOWSKI *et al.*, 2013). Segundo o mesmo estudo, um dos fatores de risco para as dores nas extremidades inferiores envolve a postura e/ou função dos pés (RISKOWSKI *et al.*, 2013).

A pronação da articulação subtalar está necessariamente associada ao aumento da rotação medial dos membros inferiores durante as atividades de suporte de peso (RESENDE *et al.*, 2015; SOUZA, THALES R *et al.*, 2010). Estudo de Pohl *et al* (2007), sugere que o principal componente de pronação da articulação subtalar no plano coronal do retropé – eversão/inversão - seja transferido para rotação do plano transverso da tíbia, onde movimentos anormais do pé podem alterar a cinética e a cinemática dos membros inferiores, resultando em aumento dos riscos de lesões nas estruturas dos tecidos moles e/ou ósseos (POHL; MESSENGER; BUCKLEY, 2007). Segundo Chuter *et al* (2012), a pronação excessiva ou prolongada pode atrasar a rotação lateral da tíbia e interromper o tempo entre a extensão do joelho e supinação do retropé (CHUTER; JANSE DE JONGE, 2012). Além disso, o aumento da eversão do calcâneo associado à pronação excessiva da articulação subtalar, poderia resultar no aumento da abdução de joelho em atividades de cadeia cinética fechada (FONSECA *et al.*, 2007; JOHANSON; HUNG; WALTERS, 2010).

Semelhante à marcha, a descida de degrau também é dividida nas fases de apoio e de balanço, e em cada fase, são divididas em subfases (GERSTLE, 2014; JE ZACHAZEWSKI, PO RILEY, 1993). A fase de apoio consiste nas seguintes subfases: duplo apoio, aceitação de peso, apoio de um único membro, continuação para a frente e descida controlada (GERSTLE, 2014; JE ZACHAZEWSKI, PO RILEY, 1993). A fase de apoio compreende 68 ± 2% do ciclo da descida de degrau total, podendo ser subdividida em 0 a 14 % na aceitação de peso, 14 a 34% em continuação para a frente e 34 a 68% na descida controlada. O apoio de um único membro representa 39% da fase de apoio (14%-53% de cada ciclo) (JE ZACHAZEWSKI, PO RILEY, 1993). O duplo apoio ocorre no início e no final da fase de apoio, 0%-14% e 53%-68%, respectivamente (JE ZACHAZEWSKI, PO RILEY, 1993). A fase de balanço compreende os 32% restante da descida de degrau e consiste na retirada do membro e no posicionamento do pé (GERSTLE, 2014; JE ZACHAZEWSKI, PO RILEY, 1993) (Quadro 1).

<table>
<tr><td>DUPLO APOIO</td><td colspan="2">APOIO DE UM ÚNICO MEMBRO</td><td>DUPLO APOIO</td><td rowspan="2">RETIRADA DO MEMBRO</td><td rowspan="2">POSICIONAMENTO DO PÉ</td></tr>
<tr><td>Aceitação de peso</td><td>Continuação para a frente</td><td colspan="2">Descida controlada</td></tr>
<tr><td colspan="4">FASE DE APOIO</td><td colspan="2">FASE DE BALANÇO</td></tr>
</table>

0 14 32 34 53 58 68 84 100

**Quadro 1**. Fases da descida de degrau. Os números ao longo da linha de base representam o ciclo percentual gasto em cada fase ou subfase de 0 a 100% das fases de descida de degrau.


**Figura 1**. Duplo apoio – subfase: aceitação de peso.

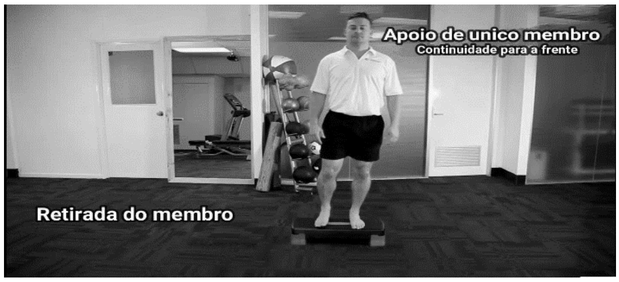


**Figura 2**. Apoio de único membro – subfases: continuidade para a frente e retirada do membro.


**Figura 3**. Apoio de único membro – subfase: descida controlada.


**Figura 4**. Duplo apoio – subfases: descida controlada e posicionamento do pé.

Lin et al (2004), realizaram uma análise tridimensional da cinética e da cinemática dos membros inferiores durante a subida e descida de degraus, e apontaram que durante a descida de degraus, no contato inicial, três articulações do membro inferior estendem para atingir o degrau inferior, seguida de flexões no joelho e tornozelo para absorção de impacto e aceitação do peso corporal (LIN; LU; HSU, 2004). As três articulações estabilizam o posicionamento do corpo durante a fase de apoio enquanto o outro membro se encontra na fase de balanço (LIN; LU; HSU, 2004).

No início da tarefa de descida de degraus, mais precisamente na subfase de continuidade para frente, os seres humanos devem transportar ativamente o centro de gravidade para frente e em seguida resistir à gravidade durante a subfase de descida controlada. Segundo estudo, essa capacidade de resistir a força imposta pela gravidade é conseguida através da contração muscular excêntrica, que controla a taxa de abaixamento do centro de gravidade absorvendo energia cinética (RICHARD *et al.*, 2010). Estudo de Lin et al (2004), observaram através de análise eletromiográfica do músculo glúteo máximo que uma ativação concêntrica do quadril foi necessária para mover o centro de massa corporal para frente no contato inicial e uma força excêntrica foi usada para desacelerar a queda do centro de massa corporal em direção ao membro contralateral na descida controlada (LIN; LU; HSU, 2004).

Durante a subfase de descida controlada, a articulação do joelho começa a partir de uma posição estendida relativamente estável e flexiona, em direção a uma posição cada vez menos estável (SELFE *et al.*, 2008). Segundo Silva *et al* (2015), durante a descida de degraus, são realizadas contrações excêntricas dos músculos reto femoral, vasto lateral, sóleo e gastrocnêmio medial, que agem contra a força da gravidade, visando o controle da flexão do joelho e minimizando o impacto do pé contra o solo durante a descida controlada (SILVA, D. D. O. *et al.*, 2015). Na ausência de forte atividade muscular excêntrica ao redor do joelho, o centro de massa aceleraria devido à gravidade (SELFE *et al.*, 2008). Além disso, o controle excêntrico dos flexores plantares de tornozelo são importantes na biomecânica do movimento de descida de degrau, pois, requer um momento flexor plantar relativamente alto devido ao contato no antepé, em vez do movimento típico com contato de calcanhar na posição inicial (ALCOCK; O'BRIEN; VANICEK, 2015; FRANÇOIS; PELLAND; ROBERTSON, 2008).

Alguns estudos analisaram a resposta eletromiográfica de músculos excêntricos durante a atividade de descida de degraus (ANDRIACCHI *et al.*, 2008; HONG; SHIN, 2015; OSKOUEI *et al.*, 2014). Estudo de Hong *et al* (2015), aponta que a modulação da amplitude eletromiográfica dos músculos reto femoral e gastrocnêmio é maior que a dos músculos semitendinoso e sóleo durante a descida de degraus (HONG; SHIN, 2015). Estudo de Oskouei *et al* (2014), investigaram a atividade eletromiográfica dos músculos sóleo e tibial anterior envolvidos na subida e descida de degraus em diferentes alturas e observaram que a atividade eletromiográfica do músculo sóleo foi significativamente maior em relação ao tibial anterior ao subir e descer degraus (OSKOUEI *et al.*, 2014). Andriacchi *et al* (2008), observaram durante a descida de degraus, uma diminuição da atividade do tibial anterior na transição do último degrau para o início da realização da marcha (ANDRIACCHI *et al.*, 2008). Segundo estudo de Willem *et al* (1995), em atividades como a marcha, o tibial anterior se contrapõe excentricamente e atua em conjunto com outros músculos na aceitação de peso, absorção de choque, desaceleração da flexão plantar e para resistir à pronação do pé (WILLEM *et al.*, 1995).

O estudo de Baldon *et al* (2013), analisaram a diferença na cinemática dos membros inferiores durante a descida de degraus entre gêneros, e foram observadas as presenças de rotação medial de joelho e valgo de joelho durante a manobra de agachamento de uma única perna, sendo ambas maiores em mulheres (BALDON *et al.*, 2013). Durante o apoio de um único membro, o abdutor do quadril e os músculos rotadores laterais são desafiados à estabilizar a articulação da pelve e do quadril nos planos coronal e transverso (BALDON *et al.*, 2013). Já o estudo de Singhal *et al* (2014) investigaram a diferença entre gêneros idosos saudáveis na descida de degrau e observaram que as mulheres apresentaram momentos de abdução do quadril maiores em comparação com os homens (SINGHAL *et al.*, 2014). Segundo o mesmo estudo, os momentos de abdução do quadril estão relacionadas ao equilíbrio do plano coronal e são necessárias durante as fases de aceitação de peso e descida controlada (SINGHAL *et al.*, 2014). Earl *et al* (2007), compararam a cinemática dos membros inferiores entre gêneros em atividades de escada e verificaram que as mulheres apresentaram maior rotação medial e valgo de joelho, bem como maior adução do quadril durante a tarefa de descida de degrau (EARL; MONTEIRO; SNYDER, 2007).

O valgismo dinâmico do joelho é a alteração biomecânica envolvendo todo o membro inferior, caracterizado pelos movimentos excessivos de adução e rotação medial de quadril, abdução do joelho e eversão do tornozelo, podendo ter relação com o aumento da pronação da articulação subtalar (DONATELLI *et al.*, 1999; ISHIDA *et al.*, 2014; TRAN *et al.*, 2016). A presença do valgismo durante os movimentos do membro inferior pode, também, alterar dinamicamente o alinhamento da patela, o que pode aumentar a sobrecarga em estruturas como os retináculos patelares, cartilagem articular e coxim adiposo e predispor o desenvolvimento de dor patelofemoral (BITTENCOURT, 2010). É pensado que o mau alinhamento do membro inferior, que resulta em mau posicionamento compensatório do pé, contribui para a dor patelofemoral (HEINTJES *et al.*, 2003). Estudos apontam que compensações proximais associadas a desalinhamentos dos pés, poderiam influenciar significativamente a mecânica da articulação patelofemoral e contribuírem para o desenvolvimento da dor patelofemoral (HEINTJES *et al.*, 2003; LEE, J.; YOON; CYNN, 2016; POWERS *et al.*, 2012).

James *et al* (1978), durante um estudo que analisava a etiologia das lesões em corredores, concluiu que a dor anterior de joelho estava relacionada à rotação anormal do plano transverso (JAMES; BATES; OSTERNIG, 1978). Neste mesmo estudo, eles observaram que as rotações no plano transverso estavam associados a pronação e supinação da articulação subtalar, onde rodava internamente com a pronação e externamente com a supinação (JAMES; BATES; OSTERNIG, 1978). No estudo de Tiberio (1987), foi observado que durante a corrida, a eversão do retropé e a flexão do joelho induzem a rotação medial da tíbia, enquanto que a inversão do retropé e a extensão do joelho induzem a rotação lateral da tíbia (TIBERIO, 1987). Segundo Kagaya *et al* (2015), a eversão do retropé é pensada para ser acoplada com a rotação medial da tíbia não apenas quando está em pé, mas também a fase de posição da marcha ou corrida (KAGAYA; FUJII; NISHIZONO, 2015). Em cada caso, uma falha estrutural faz com que o retropé faça uma eversão excessiva após o contato do calcanhar (SOUZA, THALES REZENDE DE *et al.*, 2011). James *et al* (1978), concluiu que se a torção tibial interna fosse aumentada e prolongada com a pronação excessiva da articulação subtalar, ocorreria mais rotação no joelho e maior probabilidade de dor anterior no joelho (JAMES; BATES; OSTERNIG, 1978).

Segundo Antônio *et al* (2014), marcha no plano e a marcha em degraus apresentam diferenças quanto ao padrão de contato com o pé, com a marcha no plano consistindo de contato de calcanhar enquanto a marcha de degraus é composta por um contato de ponta a ponta com os dedos (ANTONIO; PERRY, 2014). Estudos vêm apontando a posição do antepé em varo na cinemática do membro inferior (LUFLER *et al.*, 2017; SCATTONE SILVA; MACIEL; SERRÃO, 2015). O antepé varo é definido como uma deformidade estática onde o antepé está em posição de inversão em relação ao retropé, quando a articulação subtalar é fixada na posição neutra (DAVIS, 2005; SCATTONE SILVA; MACIEL; SERR??O, 2015). Durante o suporte de peso, os segmentos conectados dos membros inferiores funcionam como uma cadeia cinemática fechada, portanto, uma deformidade estrutural em varo pode levar a compensações em articulações proximais e tecidos moles (LUFLER *et al.*, 2017). Considera-se que o antepé varo induz a pronação aumentada das articulações subtalar e mediotarsal durante a postura do pé, permitindo que os metatarsos mediais entrem em contato com o chão, o que por sua vez, pode fazer com que a tíbia e o fêmur gire internamente (LUFLER *et al.*, 2017; SCATTONE SILVA *et al.*, 2013; SCATTONE SILVA; MACIEL; SERRO, 2015). Isso pode causar uma queda da maioria das estruturas do pé em direção ao seu lado medial, aumentando o ângulo tibiocalcâneo (ALONSO-VÁZQUEZ *et al.*, 2009). Em vez de uma alavanca rígida, o antepé pode se tornar uma estrutura móvel durante a propulsão da marcha, produzindo maiores forças de compressão e cisalhamento transmitidas para os tecidos moles circundantes (ALONSO-VÁZQUEZ *et al.*, 2009). Todas essas alterações podem ter efeitos negativos sobre o resto do pé, as articulações mais proximais do membro inferior e da coluna vertebral, que terão que se adaptar a essas modificações (ALONSO-VÁZQUEZ *et al.*, 2009). A pronação excessiva tem sido relacionada com a presença do alinhamento em varo do antepé, o qual pode aumentar o torque eversor do complexo tornozelo – pé após o contato do antepé com a superfície de suporte (SOUZA, THALES R. *et al.*, 2009).

Outros estudos vêm abordando as direções dos dedos dos pés – metatarsos - em relação aos ângulos de abdução e rotação do joelho (ISHIDA *et al.*, 2014; TRAN *et al.*, 2016). Em um estudo de Ishida *et al* (2014) sobre a rotação do joelho associada ao valgo dinâmico de joelho e a direção dos dedos dos pés, apontou que a postura de dedos para fora causa abdução e rotação lateral do joelho. Já a postura de dedos para dentro causa rotação medial de joelho, onde consequentemente causa diminuição da abdução do joelho(ISHIDA *et al.*, 2014). A postura dos dedos para fora e dedos para dentro induz a rotação lateral e interna do pé, respectivamente, e estas rotações por sua vez, afeta a rotação lateral e interna da tíbia através da articulação do tornozelo (ISHIDA *et al.*, 2014).

Os clínicos geralmente concordam que algumas formas de órteses ou calçados ortopédicos especializados podem controlar movimentos como da pronação excessiva (KOSONEN *et al.*, 2017; OH *et al.*, 2017). As órteses plantares ou palmilhas ortopédicas são comumente prescritas por médicos, fisioterapeutas e podólogos (KIDO *et al.*, 2014). As órteses plantares podem ser feitas sob medida, ou seja, com base em uma representação tridimensional do pé do indivíduo ou pré-fabricadas de forma genérica (HELEN R. BRANTHWAITE; CARL J. PAYTON; NACHIAPPAN CHOCKALINGAM, 2004). As Palmilhas biomecânicas são fabricadas com material termo moldável para criar uma base de suporte aos pés, onde serão adicionadas cunhas mediais, laterais, anteriores ou posteriores no interior de calçados de indivíduos com alterações estruturais (GUIMARÃES *et al.*, 2006; TELFER *et al.*, 2013). Estas cunhas tem como objetivo o controle dos movimentos de pronação e supinação dos pés desempenhados em cadeia cinética fechada durante atividades funcionais (RODRIGUES *et al.*, 2013).

A análise tridimensional do movimento permite que o pé seja avaliado em cada plano de movimento (HELEN R. BRANTHWAITE; CARL J. PAYTON; NACHIAPPAN CHOCKALINGAM, 2004). Estudos de análise tridimensional do movimento do pé com uso de órteses plantares mostrou que as órteses apresentam capacidade de afetar o retropé significativamente, reduzindo o ângulo máximo da pronação e o período total da pronação (OH *et al.*, 2017; RODRIGUES *et al.*, 2013).

Segundo Hamlyn *et al* (2012), as órteses plantares proporcionam maior controle do pé, redução de estresses biomecânicos, promove suporte aos arcos do pé, melhora a absorção de choque, aumenta as capacidades proprioceptivas e posiciona a articulação subtalar em uma posição mais mecanicamente estável (HAMLYN; DOCHERTY; KLOSSNER, 2012). Segundo estudos, as órteses plantares vêm sendo amplamente utilizadas na prática clínica dos profissionais da saúde como auxiliares no tratamento de alterações músculo esqueléticas como a pronação excessiva, eversão do pé, aumento da rotação medial da tíbia, aumento do momento de inversão do tornozelo e aumento do momento de adução e rotação lateral de joelho (CHEUNG; CHUNG; NG, 2011; HELEN R. BRANTHWAITE; CARL J. PAYTON; NACHIAPPAN CHOCKALINGAM, 2004; SOUZA, THALES REZENDE DE *et al.*, 2011), e de condições clínicas distais, como a fasceíte plantar, ou para condições proximais, como a dor patelofemoral (ANDERSON; STANEK, 2013; BOLDT *et al.*, 2013; CROSSLEY *et al.*, 2016), controlando potencialmente as movimentações excessivas ou prolongadas da articulação subtalar (NAWOCZENSKI; LUDEWIG, 2004).

O estudo de Garbalosa *et al* (2015), aponta a importância do uso de órteses plantares para o controle de alterações nos movimentos dos pés (GARBALOSA *et al.*, 2015). Segundo o estudo, durante a fase de contato inicial da marcha, o calcâneo e o tálus irão realizar movimentos de eversão, flexão plantar e adução, respectivamente, resultando em uma rotação medial da tíbia e dorsiflexão, abdução e inversão do antepé (GARBALOSA *et al.*, 2015). Se o padrão normal de movimento do pé é alterado, ocorrendo em tempo prolongado ou em quantidade excessiva, as articulações do retropé e mediopé apresentarão aumento de estresses intrínsecos, como a fáscia plantar, e extrínsecos, como a articulação patelofemoral. E esse estresse repetitivo não sendo alterados, podem levar a sintomas em indivíduos com lesões de uso excessivo da extremidade inferior (GARBALOSA *et al.*, 2015). Essas alterações podem originar compensações durante atividades funcionais como marcha, corrida e práticas esportivas, levando a disfunções e patologias (GUIMARÃES *et al.*, 2006).

As órteses plantares podem ser feitas sob medida, ou seja, com base em uma representação tridimensional do pé do indivíduo ou pré-fabricadas de forma genérica (HELEN R. BRANTHWAITE; CARL J. PAYTON; NACHIAPPAN CHOCKALINGAM, 2004). As Palmilhas biomecânicas são fabricadas com material termo moldável para criar uma base de suporte aos pés, onde serão adicionadas cunhas mediais, laterais, anteriores ou posteriores no interior de calçados de indivíduos com alterações estruturais (GUIMARÃES *et al.*, 2006; TELFER *et al.*, 2013). Estas cunhas tem como objetivo o controle dos movimentos de pronação e supinação dos pés desempenhados em cadeia cinética fechada durante atividades funcionais (RODRIGUES *et al.*, 2013). Estudo de Sasaki & Yasuda (1987), aponta que o arco em palmilhas podem transferir cargas de forma contralateral ao posicionamento do arco na articulação do joelho, ou seja, se o indivíduo faz uso de palmilha com arco longitudinal medial, cargas serão impostas em região lateral da articulação do joelho, e se o indivíduo faz uso de palmilha com arco longitudinal lateral, cargas serão impostas de forma contralateral no joelho (SASAKI; YASUDA, 1987). O alívio da dor e a capacidade de retorno aos níveis anteriores de atividades são medidas usadas para avaliar o sucesso da intervenção ortótica (NAWOCZENSKI; LUDEWIG, 2004).

Acredita-se que a pronação excessiva ou prolongada ao longo da postura limita a capacidade do primeiro metatarso e cuneiformes - primeiro raio - a fornecer uma estrutura estável para a propulsão (NAWOCZENSKI; LUDEWIG, 2004). Uma opção de design de órtese plantares para pés anormalmente pronados incorpora o uso de órteses com cunha longitudinal medial em regiões do antepé e retropé (NAWOCZENSKI; LUDEWIG, 2004). Estudo aponta que atletas que utilizavam palmilhas com 5º de elevação da borda medial do pé apresentaram uma redução do valgismo de joelho e eversão/pronação do calcâneo durante a aterrissagem do salto (JOSEPH *et al.*, 2008). A natureza do design sob o arco longitudinal medial é teorizado para promover além do controle excessivo da pronação, a primeira flexão plantar metatarsiana, restaurando assim o arco fisiológico (NAWOCZENSKI; LUDEWIG, 2004). O objetivo final de cada órtese plantar é buscar restaurar o padrão típico da marcha e sem dor, permitindo que o pé acomode as variações da superfície e proporcionam uma estabilidade adequada para a propulsão (NAWOCZENSKI; LUDEWIG, 2004).

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# CAPÍTULO 2

## Influence and benefits of foot orthoses on kinematics, kinetics and muscle activation during step descent task


## ABSTRACT

Medial wedged foot orthoses are frequently prescribed to reduce retropatellar stress in patients with patellofemoral pain (PFP) by controlling calcaneal eversion and internal rotation of the tibia. During activities of daily living, the highest patella loads occur during stair descent, but the effect of foot orthoses during stair descent remains unclear. The purpose of this study was to compare the kinematics, kinetics and muscle activation during a step descent task in healthy volunteers using three designs of foot orthoses (insoles). Sixteen healthy subjects with a mean age of 25.7 years, BMI of 23.3, and +5 Foot Posture Index were recruited. Subjects performed a step down task from 20 cm using a 5º rearfoot medial wedge (R), a 5º rearfoot and forefoot medial wedge (R/F), and a control flat insole (C). Significant improvements in control were seen in the R and R/F insoles over the C insole in the foot and at the ankle and hip kinematics. The R and R/F insoles increased the knee adduction moments, but decreased the knee internal rotation moment compared to the C insole. Adductor hallucis (AH) activity was reduced with both insoles, whereas tibialis anterior (TA) activity was reduced_with the R insole only. Foot orthoses can change joint mechanics in the foot and lower limbs providing greater stability and less work done by AH and TA muscles. This data supports the use of foot orthoses to provide functional benefits during step descent, which may benefit patients with PFP.


**Words key:** Insole, biomechanics, kinetic, kinematic, step.

## 1. Introduction

The human musculoskeletal system is challenged daily across different types and levels of terrain [1]. Stairs are commonly encountered in the workplace, at home, and in the community. Although these are rarely challenging for healthy individuals, they could be considered a difficult activity of daily living for elderly, injured or disabled persons where motor function is compromised [2].

Many studies have demonstrated significant differences between stair climbing and level walking. In particular, step descent has been shown to produce greater moments and range of motion at the knee leading to a significantly greater mechanical demand [2]. Compared to step ascent, a step descent is more challenging due to the center of mass being moved both forwards and down in a controlled lowering phase [3]. This is achieved through eccentric muscular activation, which controls the rate of lowering of the center of mass [3]. In addition, during the controlled lowering phase, the knee joint starts from a relatively stable extended position and flexes towards an increasingly unstable position. The increased joint flexion causes a progressive increase in the external flexion moment which is matched by progressively increasing eccentric muscle activity and joint forces in order to prevent collapse [3]. In healthy adults, stair descent has been shown to yield greater forces and greater peak knee abduction moments compared to stair ascent and level walking [4,5]. Differences in the foot-to-ground interface have also been observed, with a heel-to-toe contact pattern during level gait and a toe-to-heel contact during stair climbing. Specifically, the metatarsal heads accept the load followed by a lowering onto the heel [6]. An additional difference between level walking and descending stairs is linked to the vertical ground reaction force (vGRF). Both tasks show two peaks in vGRF; however, during the step descent the first vertical peak is greater than the second peak [1], and although the horizontal forces are similar for braking impulse, the propulsive force is lower during step descent [1].

Changes in movement control in the feet and lower limbs may lead to compensations during functional activities and could directly be associated with risk factors and the occurrence of injuries [7]. According to Borque et al. [8], activities that include an inclined and irregular surface such as stair descent, can demonstrate symptoms which may arise from the need to compensate for inherent instabilities in the musculoskeletal system.

Foot motions such as excessive supination and pronation are transferred proximally up the lower limb, in particular through the rearfoot torque mechanism. This can generate an increased demand on structures such as the anterior cruciate ligament and patellofemoral joint [7,9]. The use of insoles to correct foot alignment is a common conservative management approach that aims to act as a mechanical barrier against excessive patterns of movement of the foot. Medial wedged foot orthoses are often prescribed to reduce the knee and hip joint loads thought to increase retropatellar stress by reducing calcaneal eversion and tibial internal rotation [9]. Clinically, medial wedges are frequently positioned under the rearfoot acting together with the arch support. However, recent studies have suggested that both the rearfoot and forefoot influence the control of movements in excessive pronation. For example, Resende et al (2015) noted that an increase in forefoot pronation may result in increased rearfoot eversion internal hip rotation during gait [10]. In an earlier study, Monaghan et al. (2014) showed that the angle of the forefoot to the ground at forefoot contact determined the amount and duration of eversion during walking. It was noted that the rearfoot angle at rearfoot contact had no effect on the amplitude or duration of eversion during walking [11]. In addition, Rodrigues et al. (2013) demonstrated that medial wedge insoles in the forefoot and rearfoot reduced eversion and eversion velocity of the ankle joint complex in runners, with and without anterior knee pain [12]. Due to the toe-to-heel contact pattern of movement during the step descent task, postings of orthoses under the forefoot could play an important role in the control of foot pronation and associated movements of proximal joints of the lower limb.

The purpose of this study was to compare the kinematics, kinetics and muscle activation during the lowering phase of the step descent task of healthy volunteers using three designs of insoles (foot orthoses). These included: a 5$^{o}$ medial wedge insole positioned under the rearfoot, a 5$^{o}$ medial wedge insole positioned under the rearfoot and forefoot, and a control flat insole.

## 2. Methods

### 2.1 Participants

Sixteen healthy subjects (10 males and 6 females) with a mean age of 25.7 years (SD 5.8), body weight of 71.7 kg (SD 10.6), height of 174.8 cm (SD 9.2), mean BMI of 23.3 (SD 1.7) and a mean score of +5 (SD 4) for the Foot Posture Index (FPI) (version 6) were recruited. All participants were free of previous and present history of patellofemoral pain, injuries to the lower-limbs or pelvis or surgery. The volunteers signed an informed consent in accordance with the Declaration of Helsinki. This study was approved by the Ethical Committee of the University of Central Lancashire.

### 2.2 Procedures

An initial assessment was conducted which included measures of body weight, height and Foot Posture Index (version 6). Lower limb kinematic data were then obtained using a 10-camera Oqus 7 system at 100 Hz (Qualisys Medical AB, Gothenburg, Sweden). Passive retro-reflective markers were placed on the lower limbs and pelvis using the Calibrated Anatomical System Technique allowing the segmental kinematics to be tracked in 6-degrees of freedom [13]. Anatomical markers were positioned by the same researcher on the anterior superior iliac spine, posterior superior iliac spine, greater trochanter, medial and lateral femoral epicondyle, medial and lateral malleoli and over medial and lateral aspects of 1$^{st}$ and 5$^{th}$ metatarsal respectively. Additionally clusters of non-collinear markers were attached to the shank and thigh, and markers were also placed over rearfoot, midfoot and forefoot aspects of the shoes [13]. Static calibration trials were obtained with the participant in the anatomical position. Kinetic data were collected using two AMTI force plates at 2000 Hz (Advanced Mechanical Technology Inc, Watertown, MA). Joint moment data were calculated using three-dimensional inverse dynamics, and the external joint moment data were normalised to body mass (Nm/kg).

In addition, electromyographic (EMG) data were obtained from the tibialis anterior (TA), peroneus longus (PL), medial gastrocnemius (MG) and abductor hallucis muscles (AH) using a Trigno Wireless EMG system at 2000 Hz (Delsys Inc., Boston, MA). The skin was cleaned with alcohol wipes and the standard EMG electrodes were positioned in accordance with the SENIAM guidelines and fixed with double-sided adhesive skin interfaces. Standard Trigno wireless sensors were used to collect data from TA, PL and MG muscles, and a Trigno Mini wireless sensor was placed such that the data from AH could be collected inside the footwear with minimal sensory disturbance (Figure 1). The AH was palpated, and the mini electrodes were fixed with a double-sided adhesive skin interface and the connecting wire was secured with Hyperfix.

**Figure 1.** Trigno Mini wireless sensor placed on the foot for abductor hallucis muscle.

Data were collected from the dominant lower limb and pelvis, with the dominant limb being defined as the limb with which they would kick a ball. Six repetitions of a 20 step descent at a self-selected speed were performed from a step positioned on the first force plate, which has been previously been used to assess closed chain eccentric control and stability [3], were performed under three randomized conditions. The conditions included: control flat insole (C-insoles), a medial longitudinal arch support with a 5 degree medial rearfoot posting insole (R insoles), and a medial longitudinal arch support with a 5 degree medial forefoot and rearfoot posting (R/F insoles). The base of the insoles was pre-fabricated with a standardised arch support and neutral heel. The 5 degree posting material was made from ethylene vinyl acetate (EVA) and was affixed under insoles using double side tape (Figure 2). All volunteers wore appropriately sized standardized footwear (Dr Comfort Winner Plus). The size of the insoles was adjusted to fit the footwear, however the insoles were not customized for each volunteer.

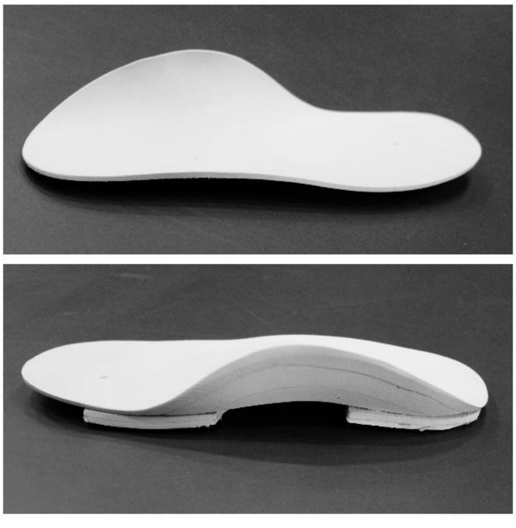

**Figure 2.** Intervention insole with arch support and 5° wedges posted under the rearfoot and forefoot.

### 2.4 Data processing

Raw kinematic, kinetic and EMG data were exported to Visual3D (C-Motion Inc., Germantown, USA). Kinematic and kinetic data were filtered using fourth order Butterworth filters with cut off frequencies of 6Hz and 25Hz, respectively. EMG data were zeroed, band-pass filtered with corner frequencies of 20Hz and 500 Hz, full-wave rectified and enveloped using a fourth-order low-pass Butterworth filter with a cut-off frequency of 25Hz. The EMG data were normalized to the maximal observed signal during the dynamic contraction during the movement tasks [6]. For all data, time was normalized to 101 points from toe-off of the non-dominant foot, using the threshold of the 1st metatarsal marker vertical trajectory velocity, to initial contact of the contralateral limb using a threshold of 10N on vertical force on the second force plate.

The mean kinematic, kinetic and EMG values for each condition were recorded during the single limb descent phase. The dependent kinematic variables included: the minimum, maximum and range of motion in the sagittal, coronal and transverse plane at the forefoot, midfoot, ankle, knee and hip. The dependent kinetic variables included: peak ankle, knee and hip moments in all three planes. Finally, the dependent variables for the peak and integrated EMG (iEMG) values from TA, PL, MG and AH were found. These were then normalized to the maximal observed signal during single limb descent phase for each muscle [14]

### 2.5 Data Analysis

Each kinematic and kinetic variable was assessed and found to be normally distributed and suitable for parametric statistical testing. Repeated measure ANOVAs with pairwise comparisons were performed to compare the three insole conditions, in addition the effect size ($p\eta^2$) was also found. Bonferroni corrections were employed to allow for multiple comparisons and to reduce the possibility of type I errors. All statistical calculations were conducted using SPSS v.22.0 (SPSS Inc., Chicago, USA), with the $\alpha$ level set at 0.05.

## 3. Results

Descriptive statistics are presented in Table 1 and time series curves for each variable with a statistically significant difference are presented in Figure 3. The Repeated Measured ANOVAs showed significant differences between the conditions during the controlled lowering phase of the step descent task, for kinematics at the foot, ankle and hip (Table 1). Further pairwise comparisons showed significant differences between conditions for the foot, ankle and hip (Table 2). At the foot and ankle, significantly less metatarsocalcaneal internal rotation (p=0.001, p=0.029), and calcaneal eversion (p=0.006, p=0.014), were seen with the R and R/F insoles compared to the C. In addition, ankle abduction was also significantly reduced using the R and R/F insoles compared to the C insole (p=0.018, p=0.007). At the hip, initiation of motion occurred with significantly less peak hip external rotation for the C insoles compared to the R and R/F insoles (p=0.007, p=0.001). All insoles showed statistically different results for hip internal rotation (p=0.017, p=0.002, P=0.023), with the C insoles showing the greatest rotation. In the coronal plane the R insole produced a significant lower hip adduction (p<0.001), and lower hip coronal plane range of motion compared to C and R/F insoles (p=0.001, p=0.007).

The repeated measures ANOVAs also showed differences in the knee moments and iEMG activity for the TA and AH between the conditions (Table 1). Further pairwise comparisons showed significantly lower knee adduction moment for the C insoles compared to the R and R/F insoles of the step descent task (p<0.001, p<0.001). However, the C insoles showed a greater knee internal rotation moment in relation to the R and R/F insoles (p<0.001, p<0.001). The iEMG data showed a higher peak of activity and iEMG in AH with the C insoles compared to the R and R/F insoles (p=0.006, p=0.014), and lower TA iEMG with the R insoles compared to the R/F and C insoles (<0.001, 0.009) (Table 3).

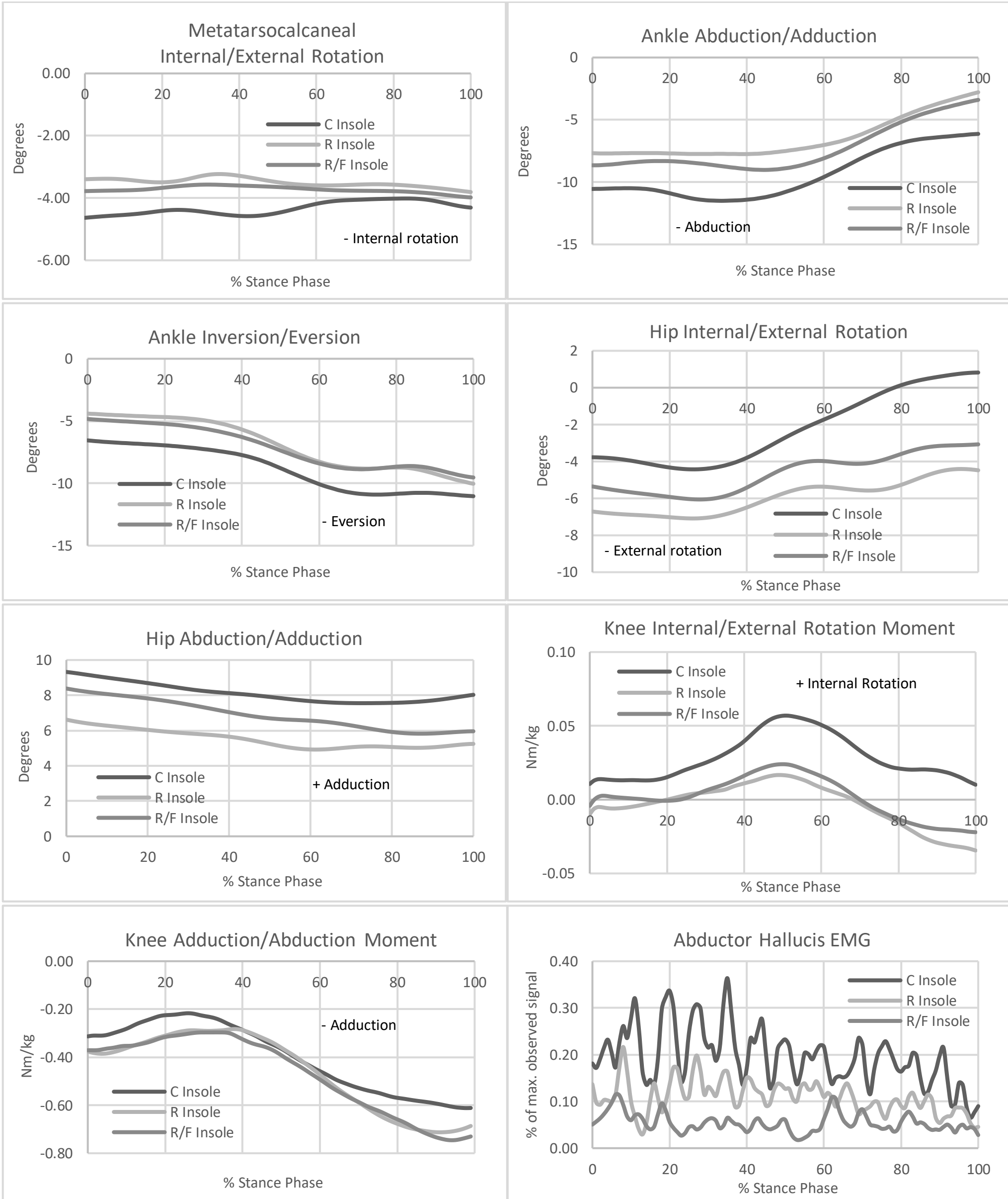


**Figure 3.** Mean Kinematic, kinetic and muscle activation parameters from a single subject.

| | R/F insoles Mean (sd) | C insoles Mean (sd) | R insoles Mean (sd) | p-value | $p\eta^2$ |
|---|---|---|---|---|---|
| **Kinematics (degrees)** | | | | | |
| Peak Metatarsal to Calcaneal Int. Rot. | -5.8 (3.3) | -6.4 (3.4) | -5.5 (3.2) | 0.002 | 0.34 |
| Peak Ankle Abduction | -11.0 (5.1) | -13.6 (5.6) | -11.2 (4.2) | 0.003 | 0.32 |
| Peak Ankle Eversion | -10.4 (5.7) | -11.3 (5.3) | -10.2 (5.3) | 0.002 | 0.33 |
| Hip ROM in the coronal plane | 5.9 (2.4) | 6.3 (2.3) | 4.9 (2.0) | 0.001 | 0.36 |
| Peak Hip External Rotation | -6.5 (9.0) | -5.1 (9.5) | -6.8 (8.3) | 0.001 | 0.36 |
| Peak Hip Internal Rotation | 0.6 (9.1) | 1.9 (9.9) | -0.2 (8.7) | 0.001 | 0.39 |
| Peak Hip Adduction | 10.0 (3.5) | 10.9 (3.0) | 9.1 (3.1) | 0.001 | 0.39 |
| **Kinetics (Nm/kg)** | | | | | |
| Peak Knee Internal Rotation Moment | 0.013 (0.022) | 0.044 (0.034) | 0.011 (0.021) | <0.001 | 0.67 |
| Peak Knee Adduction Moment | -0.449 (0.127) | -0.388 (0.112) | -0.447 (0.131) | <0.001 | 0.64 |
| **EMG (% max. observed signal)** | | | | | |
| Peak Abductor Hallucis activity | 55.8 (16.8) | 69.5 (18.9) | 54.5 (23.3) | 0.003 | 0.32 |
| Tibialis Anterior iEMG | 60.9 (16.3) | 58.1 (18.8) | 47.8 (13.8) | 0.003 | 0.33 |
| Abductor Hallucis iEMG | 55.1 (18.6) | 73.0 (15.8) | 53.2 (21.1) | 0.002 | 0.34 |

Table 1: Means and standard deviations for the kinematic, kinetic and muscle activation parameters during the lowering phase of the step descent task under the different orthotic conditions.

| | Comparisons | Mean Differences | p.value | Standard Error | Lower Bound | Upper Bound |
|---|---|---|---|---|---|---|
| **Peak Metatarsocalcaneal Internal Rotation** | R/F to C insoles | 0.6 | 0.029* | 0.26 | 0.07 | 1.17 |
| | R/F to R insoles | -0.3 | 0.201 | 0.21 | -0.71 | 0.16 |
| | R to C insoles | 0.9 | 0.001* | 0.23 | 0.40 | 1.39 |
| **Peak Ankle Abduction** | R/F to C insoles | 2.6 | 0.007* | 0.83 | 0.82 | 4.35 |
| | R/F to R insoles | 0.1 | 0.801 | 0.55 | -1.03 | 1.32 |
| | R to C insoles | 2.4 | 0.018* | 0.92 | 0.48 | 4.40 |
| **Peak Ankle Eversion** | R/F to C insoles | 0.9 | 0.014* | 0.31 | 0.20 | 1.53 |
| | R/F to R insoles | -0.3 | 0.317 | 0.24 | -0.76 | 0.26 |
| | R to C insoles | 1.1 | 0.006* | 0.35 | 0.37 | 1.87 |
| **Hip ROM in the Coronal Plane** | R/F to C insoles | -0.4 | 0.332 | 0.39 | -1.21 | 0.44 |
| | R/F to R insoles | 1.0 | 0.007* | 0.33 | 0.32 | 1.75 |
| | R to C insoles | -1.4 | 0.001* | 0.35 | -2.16 | -0.68 |
| **Peak Hip External Rotation** | R/F to C insoles | -1.4 | 0.001* | 0.38 | -2.25 | -0.63 |
| | R/F to R insoles | 0.3 | 0.433 | 0.38 | -0.50 | 1.12 |
| | R to C insoles | -1.7 | 0.007* | 0.56 | -2.94 | -0.54 |
| **Peak Hip Internal Rotation** | R/F to C insoles | -1.3 | 0.023* | 0.53 | -2.48 | -0.21 |
| | R/F to R insoles | 0.7 | 0.017* | 0.02 | 0.15 | 1.35 |
| | R to C insoles | -2.1 | 0.002* | 0.00 | -3.33 | -0.87 |
| **Peak Hip Adduction** | R/F to C insoles | -0.9 | 0.061 | 0.43 | -1.80 | 0.04 |
| | R/F to R insoles | 0.9 | 0.042* | 0.40 | 0.03 | 1.72 |
| | R to C insoles | -1.8 | <0.001* | 0.38 | -2.56 | -0.95 |

* The mean difference is significant at the 0.05 level.

Table 2: Pairwise comparisons for foot, ankle and hip kinematics between the orthotic conditions during the lowering phase of the step descent task

| | Comparisons | Mean Differences | p.value | Standard Error | Lower Bound | Upper Bound |
|---|---|---|---|---|---|---|
| **Peak Knee Internal Rotation Moment** | R/F to C insoles | 0.031 | <0.001* | 0.01 | -0.04 | -0.02 |
| | R/F to R insoles | 0.003 | 0.364 | 0.00 | 0.00 | 0.01 |
| | R to C insoles | 0.034 | <0.001* | 0.01 | -0.05 | -0.02 |
| **Peak Knee Adduction Moment** | R/F to C insoles | -0.061 | <0.001* | 0.01 | -0.02 | -0.03 |
| | R/F to R insoles | 0.003 | 0.354 | 0.01 | -0.02 | 0.01 |
| | R to C insoles | -0.058 | <0.001* | 0.01 | -0.07 | -0.03 |
| **Peak Abductor Hallucis activity** | R/F to C insoles | -13.7 | 0.013* | 4,901 | -24.20 | -3.30 |
| | R/F to R insoles | 1.3 | 0.777 | 4,591 | -8.46 | 11.11 |
| | R to C insoles | -15.1 | 0.001* | 3,848 | -23.27 | -6.87 |
| **Tibialis Anterior iEMG** | R/F to C insoles | 2.8 | 0.553 | 4,658 | -7.10 | 12.75 |
| | R/F to R insoles | 13.1 | <0.001* | 2,354 | 8.05 | 18.09 |
| | R to C insoles | -10.2 | 0.009* | 3,454 | -17.61 | -2.89 |
| **Abductor Hallucis iEMG** | R/F to C insoles | -17.8 | 0.014* | 6,458 | -31.61 | -4.08 |
| | R/F to R insoles | 1.9 | 0.633 | 3,926 | -6.45 | 10.28 |
| | R to C insoles | -19.8 | 0.006* | 6,100 | -32.76 | -6.76 |

* The mean difference is significant at the 0.05 level.

Table 3: Pairwise comparisons for kinetics and EMG between the orthotic conditions during the lowering phase of the step descent task

## 4. Discussion

This study compared the kinematics, kinetics and muscle activation during the lowering phase of the step descent task of healthy volunteers using three designs of insoles. The results showed a significant reduction in pronation of the foot and associated coupled movements on lower limbs wearing the R and R/F insole compared to the C insole.

During closed chain activities, joint motions of the lower limb are interdependent, and excessive movements from one joint may overload tissues in the kinematic chain [15]. Studies have focused on the clinical relevance of excessive pronation during walking, which has been identified as a major factor in the development of overuse injuries and PFP [12,16]. This is usually associated with a lack of muscle strength, stability, and overuse of the foot muscles due to the oblique angle of the subtalar joint [10,17]. Insoles (foot orthoses) are a common modality employed by many clinicians with evidence indicating that they can prevent overuse conditions of lower limbs [18]. Whilst the mechanism of action is debated, Hamlyn et al. previously noted that insoles with MLA support increased the area of contact under the feet reducing excessive pronation in individuals with functional ankle instability [19]. Redmond at al. (2008) showed that a slightly pronated foot posture, mean FPI of +4 (SD 3), is the normal position at rest [20]. Although, in this current study the foot posture was assessed and found to be +5 with a standard deviation of 4, indicating that on average the sample did not present with excessive pronation [20]. However, the results did show significant reductions in metatarsal to calcaneal internal rotation, ankle eversion and ankle abduction in the R and R/F compared to the C insole. This indicates that the insoles with the medial longitudinal arch (MLA) support offered greater control of the movements commonly associated with excessive pronation, and reduced work done by the AH and TA muscles. These muscles have been shown to play a role in stabilizing the MLA and consequently the pronation of the foot, which has been linked to an increase in load and MLA compression [21]. During walking, the TA's main action is eccentric control of the foot during weight acceptance, and potentially influencing the rate of rearfoot eversion [22,23]. Theoretically, forefoot motion will also be influenced by the TA, as its actions are described as elevation and lateral rotation of the 1$^{st}$ metatarsal and medial cuneiform, as well as raising the MLA during push-off [22]. Although Cornwall et al. considered the TA to be active from before the foot strikes the ground until the foot is flat [23], during a step descent the TA activity, lower limb kinematics and kinetics data show a consistently higher coefficient of variation compared to walking gait [24].

iEMG of the AH muscle has been studied by Reeser et al. who demonstrated significant myoelectric activity during late stance and the toe-off phase of gait [25]. To date, no studies have analyzed AH muscle activity during step descent. This study showed significantly lower TA and AH activity with greater joint control when using the R and R/F insoles compared to the C insole during the step descent. This may indicate better foot position control during step descent and improved support by the insoles and requires further study.

Pierrynowski showed that female participants with PFP descended with the hip more adducted and internally rotated compared to asymptomatic individuals during a step-down activity [26]. Evidence indicates that a relationship exists between excessive eversion of the rearfoot and PFP [27,28]. From a theoretical perspective, excessive eversion of the rearfoot causes excessive internal rotation of the tibia, which consequently creates a higher internal rotation and adduction of the hip. Adduction and internal rotation of the hip are reported to increase dynamic knee valgus and lateral tracking of the patella, which can lead to a reduction in the contact area of the patella, which increases PFP and stress [15]. This study observed a reduction in internal rotation and hip adduction with R and R/F insoles compared to the C insole. The R insole showed a smaller reduction compared to the R/F insole and a lower hip range of motion (coronal plane), however it appears that R and R/F insoles can influence the control of excessive movement being transferred proximally.

The knee joint plays an important role in controlling the movement and during step descent [13,29] with the hip and ankle indirectly influencing knee kinematics [10]. Previous studies have shown that dynamic knee valgus can predispose lower limb injuries, especially at the knee, where step descent activity may be an important clinical outcome. The presence of alterations in movement patterns, distally and proximally, can generate overload in the lower limb joints contributing to musculoskeletal pain [30]. In this current study, R and R/F insoles significantly decreased internal rotation and knee adduction compared to the C insole which has the potential to benefit individuals who have PFP. Whilst the findings of this study showed that F and R/F insoles can induce significant functional improvements, further evaluation is need on individuals with PFP pain and in individuals with different foot postures. In addition, this current study has reported on a 20 cm stepdown task, rather than continuous stair descent, which may yield different results [24].

Foot orthoses can change joint mechanics in the foot and lower limbs providing greater stability and less work done by AH and TA muscles. This data supports the use of foot orthoses to provide functional benefits during step descent, which may benefit patients with PFP. It should be noted however, that these results were obtained from healthy volunteer participants and this work must be replicated in patients diagnosed with PFP.

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## CAPÍTULO 3 – Considerações finais

O presente estudo demonstrou que durante a tarefa de descida de degrau, as palmilhas de arco longitudinal medial com cunhas de 5º no antepé e retropé e apenas no retropé, apresentaram resultados significativos quanto a cinemática, cinética e ativação muscular de membros inferiores e pelve. Esse resultado tem uma importância clínica, pois, estudos foram apontados na presente revisão do aumento de queixas álgicas durante a tarefa de descida de degrau, principalmente em indivíduos com dor patelofemoral.

Tendo em vista que o presente estudo foi realizado com indivíduos assintomáticos, novos estudos com indivíduos sintomáticos devem ser replicados, com a hipótese de que o uso de palmilhas com cunha medial durante a tarefa de descida de degrau, poderá altera a mecânica articular do pé e dos membros inferiores, proporcionando maior estabilidade e diminuição de queixas álgicas.

Clinicamente, novos estudos poderá fornecer subsídios para que profissionais da área de saúde possam ter em mãos uma conduta que possa complementar no processo de tratamento de indivíduos que apresentem queixas durante a tarefa de descida de degrau.

**ANEXO I – Termo de Consentimento Livre e Esclarecido (TCLE)**

**Version 1 - 08/07/15**

# CONSENT FORM

**Title of Project: Effect of insoles with medial wedge on the kinematics and kinetics of lower limbs of patients during gait and step descent**

Name of Researchers: Professor Jim Richards, Professor James Selfe, Dr Renato Trede

The following test will require you to have various markers attached to your body in order to model the way in which you walk and down step. This will also prevent you from being identified in any report/publication.

The procedure should cause you no discomfort, however, if you do feel some discomfort attempts will be made to remedy the situation.

Before any of the tests are conducted the institutional review board require written consent, please complete if you agree to the terms of the research.

**Please initial box**

1. I confirm that I have read and understand the information sheet (version 1) for the above study and have had the opportunity to ask questions. ☐
2. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving any reason, without my rights being affected. ☐
3. I agree to take part in the above study. ☐

| Name of Patient | Date | Signature |
|---|---|---|
| Name of Patient | Date | Signature |
| Name of Patient | Date | Signature |

1 for participant; 1 for researcher

## ANEXO II – Aprovação do Comitê de Ética

University of Central Lancashire

11th September 2015

James Richards/James Selfe

School of Sport and Wellbeing

University of Central Lancashire

Dear James/James,

**Re: STEMH Ethics Committee Application Unique Reference Number: STEMH 385**

The STEMH ethics committee has granted approval of your proposal application 'Effect of insoles with medial wedge on the kinematics and kinetics of lower limbs of patients during gait and step descent'.
Approval is granted up to the end of project date* or for 5 years from the date of this letter, whichever is the longer.

It is your responsibility to ensure that

- the project is carried out in line with the information provided in the forms you have submitted
- you regularly re-consider the ethical issues that may be raised in generating and analysing your data
- any proposed amendments/changes to the project are raised with, and approved, by Committee
- you notify roffice@uclan.ac.uk if the end date changes or the project does not start
- serious adverse events that occur from the project are reported to Committee
- a closure report is submitted to complete the ethics governance procedures (Existing paperwork can be used for this purposes e.g. funder's end of grant report; abstract for student award or NRES final report. If none of these are available use e-Ethics Closure Report Proforma).

Yours sincerely,

Paola Dey

Deputy Vice Chair

**STEMH Ethics Committee**

* for research degree students this will be the final lapse date

*NB - Ethical approval is contingent on any health and safety checklists having been completed, and necessary approvals as a result of gained.*

**ANEXO III – Regras de submissão Gait & Posture**

# GAIT & POSTURE

Official Journal of: Gait and Clinical Movement Analysis Society (GCMAS), European Society of Movement Analysis in Adults and Children (ESMAC), Società Italiana di Analisi del Movimento in Clinica (SIAMOC), and the International Society for Posture and Gait Research (ISPGR)

**AUTHOR INFORMATION PACK**

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**ISSN:** 0966-6362

## DESCRIPTION

*Gait & Posture* is a vehicle for the publication of up-to-date basic and clinical research on all aspects of **locomotion** and **balance**.

The topics covered include: Techniques for the measurement of **gait** and **posture**, and the standardization of results presentation; Studies of normal and **pathological gait**; Treatment of gait and **postural abnormalities**; Biomechanical and theoretical approaches to gait and posture; Mathematical models of **joint** and **muscle mechanics**; **Neurological** and **musculoskeletal** function in gait and posture; The evolution of **upright posture** and **bipedal locomotion**; Adaptations of carrying loads, walking on uneven surfaces, climbing stairs etc; spinal biomechanics only if they are directly related to gait and/or posture and are of general interest to our readers; The effect of aging and development on gait and posture; Psychological and cultural aspects of gait; Patient education.

Index bound in last issue of year.

For details of the GCMAS, ESMAC, SIAMOC, ISPGR please visit their web sites throughthese links.

## AUDIENCE

Orthopaedic surgeons, neurologists, rheumatologists, podiatrists/chiropodists, physiatrists, physical and occupational therapists, research professionals, psychologists, physiologists, bioengineers, kinesiologists, ergonomists and those with an interest in elite performance.

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[1] J. van der Geer, J.A.J. Hanraads, R.A. Lupton, The art of writing a scientific article, J. Sci. Commun.163 (2010) 51–59.

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[2] W. Strunk Jr., E.B. White, The Elements of Style, fourth ed., Longman, New York, 2000.

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[3] G.R. Mettam, L.B. Adams, How to prepare an electronic version of your article, in: B.S. Jones, R.Z.Smith (Eds.), Introduction to the Electronic Age, E-Publishing Inc., New York, 2009, pp. 281–304.

Reference to a website:

[4] Cancer Research UK, Cancer statistics reports for the UK. http://www.cancerresearchuk.org/aboutcancer/statistics/cancerstatsreport/, 2003 (accessed 13 March 2003).

Reference to a dataset:

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