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Les changements de direction : analyse biomécanique des déterminants de la performance et des facteurs de risque de blessure

Guillaume Mornieux

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Ecole Doctorale BioSE (Biologie, Santé, Environnement)

Mémoire présenté pour la candidature au diplôme

D'HABILITATION À DIRIGER DES RECHERCHES

par

Guillaume MORNIEUX

**Les changements de direction : analyse
biomécanique des déterminants de la performance
et des facteurs de risque de blessure**

(Discipline CNU 74)

Le 18 décembre 2018

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Après un travail de thèse mené entre la France et l'Allemagne et finalisé en 2005, j'ai eu la chance de partir en post-doctorat à Vancouver (Canada), à l'Université de Colombie Britannique. Travailler avec le Professeur Sanderson sur des aspects de la biomécanique du cyclisme dans la continuité de mon doctorat a été très enrichissant. Cette nouvelle expérience à l'étranger, après l'Allemagne, m'a motivé à poursuivre ma carrière à l'étranger... et revenir à l'Université de Freiburg en Allemagne pour un post-doctorat sous la direction du Professeur **Albert Gollhofer**. Trois ans après ce début de post-doctorat et ses missions de prestations de service pour Adidas, l'équipe nationale de saut à ski et autres, le Professeur Gollhofer m'a orienté vers des travaux d'HDR. Même si j'ai entre temps changé d'Université, travailler à l'Université de Freiburg, à l'Institut für Sport und Sportwissenschaft, m'aura permis de valoriser mon travail de doctorat il y a quelques temps, puis de produire le cœur de cette HDR plus récemment. Merci Albert de m'avoir accueilli (à plusieurs reprises) dans ton équipe, de m'avoir guidé dans mes différents travaux, et d'être aujourd'hui mon parrain d'HDR.

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I. BILAN DES ACTIVITES DE RECHERCHE

I.1) Thèse de Doctorat

Nouvelles méthodes de mesure de l'efficacité de pédalage : application à l'étude des facteurs biomécaniques du rendement musculaire en cyclisme (Mornieux, 2005).

Ce travail de thèse a permis de développer et valider de nouvelles méthodes de mesure de l'efficacité de pédalage afin de pouvoir analyser les facteurs biomécaniques du rendement musculaire en cyclisme.

Une première étude (Mornieux, Zameziati, Mutter, Bonnefoy, & Belli, 2006) a proposé une nouvelle méthode de mesure de l'efficacité de pédalage se basant sur la mesure des forces appliquées sur les pédales à partir de la fixation d'un ergo-cycle sur une plateforme de forces. Cette méthode, validée en conditions statique et dynamique, permet une mesure des forces en 3D appliquées sur les pédales en conservant les propriétés mécaniques des pédales. Une seconde méthodologie de mesure de l'efficacité de pédalage, se basant sur la mesure des forces tangentielle et radiale appliquées sur les pédales selon le principe de Hall, a été validée (Stapelfeldt, Mornieux, Oberheim, Belli, & Gollhofer, 2007). Elle présente comme principaux avantages, de permettre la mesure des forces appliquées sur les pédales en laboratoire et sur le terrain, de permettre la fixation de tous les types de pédale et enfin de fournir une mesure simple et valide de la puissance.

Ces méthodes de mesure des forces exercées sur les pédales ont permis ensuite de déterminer l'efficacité de pédalage et de mettre en avant sa relation avec le rendement musculaire (Mornieux, Zameziati, Rouffet, & Belli, 2006). Les différences inter-individuelles d'efficacité de pédalage, notamment lorsque celles-ci sont calculées sur la phase descendante de la pédale, expliquent les variations de rendement musculaire observées entre les cyclistes à une puissance et une cadence de pédalage données. Enfin, une dernière étude a tout d'abord démontré que l'utilisation de pédales automatiques n'avait pas d'influence sur la technique de pédalage et sur le rendement musculaire en conditions sous-maximales (Mornieux, Stapelfeldt, Gollhofer, & Belli, 2008). Ensuite, un feedback visuel du niveau de force efficace appliquée sur la pédale a été utilisé afin de contrôler l'action de traction sur la pédale en phase ascendante. Ce feedback a permis aux cyclistes loisirs d'améliorer leur efficacité de pédalage sans nuire à leur rendement musculaire. Malheureusement, tirer sur la pédale altère le rendement musculaire chez des cyclistes élités, dont le pattern de pédalage est trop ancré et donc trop sensible aux variations.

En conclusion, l'utilisation de nouvelles méthodes de mesure des forces appliquées sur les pédales a permis de mettre en évidence que l'efficacité de pédalage est un facteur mécanique de la performance, étant donné qu'elle influence le rendement musculaire. De plus, si le lien pied-pédale ne semble pas influencer la performance, la technique du feedback offre des perspectives de recherche étant donné son impact sur la performance chez les cyclistes loisirs.

I.2) Les grands axes d'évolution depuis la thèse

I.2.1) Etudes postdoctorales sur la biomécanique du cyclisme au Canada et en Allemagne

Dans la continuité de ce travail de thèse, une étude, menée à l'Université de Colombie Britannique à Vancouver au Canada, a permis d'évaluer la robustesse du pattern de pédalage dans différentes conditions mécaniques et physiologiques (Mornieux, Guenette, Sheel, & Sanderson, 2007). Pour ce faire, l'efficacité de pédalage et la répartition de la production des moments articulaires du membre inférieur ont dans un premier temps été déterminés lors d'un exercice de pédalage à différentes cadences (60, 80 et 100rpm) et puissances (150, 250 et 350W). Dans un second temps, l'influence de l'hypoxie (15% O₂) sur le pattern de pédalage a été analysée. Les différences de répartition de moment de force au niveau des différentes articulations lorsque la cadence et la puissance étaient manipulées se sont avérées faibles, voire absente au niveau de la cheville.

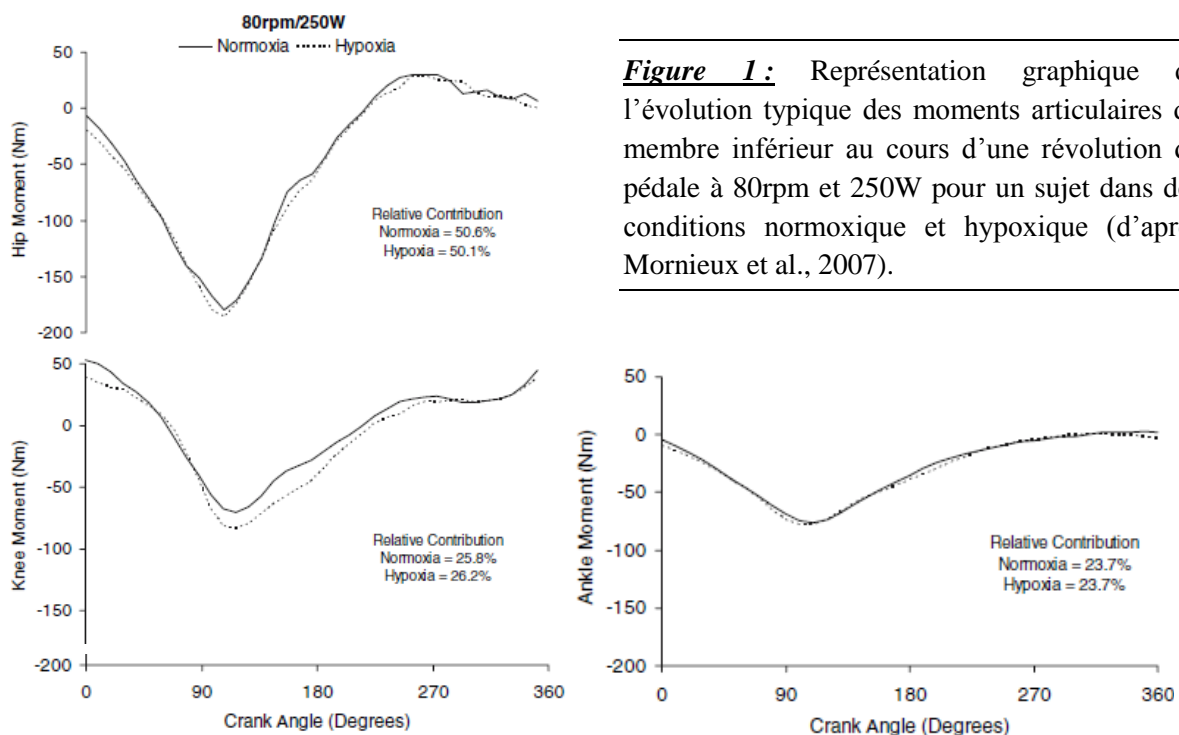


Figure 1: Représentation graphique de l'évolution typique des moments articulaires du membre inférieur au cours d'une révolution de pédale à 80rpm et 250W pour un sujet dans des conditions normoxique et hypoxique (d'après Mornieux et al., 2007).

Malgré une augmentation significative du débit ventilatoire (28%) et de la fréquence cardiaque (5%), ainsi qu'une baisse significative de la saturation en oxygène (8%), le pattern de pédalage n'était pas modifié en condition hypoxique (Figure 1). Ainsi, la coordination lors d'une tâche de pédalage demeure robuste même dans des conditions mécaniques et physiologiques extrêmes.

Par la suite, deux études ont été menées à l'Université Albert-Ludwigs de Freiburg en Allemagne. L'activation des muscles du membre inférieur lors de la phase de retour de pédale a été analysée en détail chez 9 cyclistes (Mornieux, Gollhofer, & Stapelfeldt, 2010). Cette phase avait été identifiée comme pertinente durant le travail de thèse, notamment grâce à l'ajout d'un feedback de la force efficace, permettant de travailler l'action de traction sur la pédale. Si cette action de traction permet bel et bien d'améliorer l'efficacité de pédalage par rapport à une condition de pédalage classique, la coordination neuromusculaire associée à ce nouveau pattern met en évidence une augmentation significative de l'activité électromyographique (EMG) des fléchisseurs dorsaux de l'articulation de la cheville (tibialis anterior : +69%) et des fléchisseurs du genou (biceps femoris : +48%) durant la phase de retour de pédale (Figure 2).

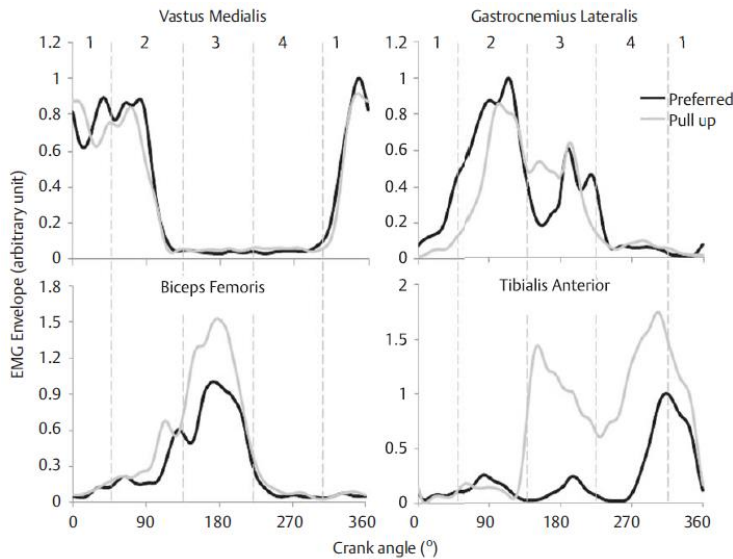


Figure 2 : Profils EMG typiques durant une révolution de pédale (0°-360°) lors d'un pédalage classique (noir) et d'un pattern de tirage sur la pédale (gris). L'activation du biceps femoris autour du point mort bas (180°) et du tibialis anterior durant la phase de retour de la pédale (180°-360°) était augmentée lors du pattern de tirage (d'après Mornieux et al., 2010).

Cependant, cette augmentation n'était pas compensée par une réduction de l'EMG des extenseurs de cheville (gastrocnemius lateralis) et de genou (vastus medialis) durant la phase descendante de la pédale, expliquant l'augmentation de consommation d'oxygène et permettant de supposer une apparition avancée de la fatigue en condition de traction sur la pédale. Ainsi, même si ce nouveau pattern de pédalage permet une amélioration de l'efficacité de pédalage, il est également associé une augmentation de l'activation neuromusculaire des

fléchisseurs de jambe et de la consommation d'oxygène, soit une probable détérioration de la performance. Ce type d'effets à court-terme ne permet cependant pas de juger complètement de la pertinence d'une nouvelle technique de pédalage sur la performance en cyclisme, étant donné le temps d'adaptation nécessaire du cycliste à ce nouveau pattern moteur.

Ainsi, afin de permettre cette adaptation, 21 cyclistes amateurs ont suivi un entraînement de quatre semaines avec au total 12 sessions d'entraînement de 30min à 60% de leur Puissance Maximale Aérobie et à 80rpm (Mornieux & Stapelfeldt, 2012). Un groupe « Contrôle » a réalisé cet entraînement sans consigne particulière, pendant qu'un groupe « Feedback » se voyait projeter en temps réel la force efficace de manière à pouvoir modifier son pattern de pédalage en tirant sur la pédale lors de la phase de retour. L'évaluation de la performance s'est faite lors d'un test d'effort incrémental avant et après période entraînement, durant lequel les forces appliquées sur les pédales étaient enregistrées sans être utilisées comme feedback pour les cyclistes. Les paramètres de puissance maximale (Pmax), fréquence cardiaque maximale (FCmax), ainsi que la puissance et la fréquence cardiaque au seuil individuel anaérobie (PiAT et FCiAT, respectivement) ont été évalués (Figure 3). Les résultats de cette étude montrent que les cyclistes ayant bénéficiés de l'entraînement spécifique avec feedback ont modifié leur pattern de pédalage, leur permettant d'être plus efficace d'un point de vue de l'application des forces sur les pédales, et ce même en l'absence de feedback. Cependant, même si une action de traction sur la pédale présenterait un avantage mécanique, l'intérêt d'un point de vue de la performance reste à être démontré puisque cette dernière évoluait de la même manière entre le groupe ayant suivi l'entraînement spécifique et le groupe contrôle (e.g. Pmax de +4%).

	Feedback		Contrôle	
	Pré	Post	Pré	Post
Pmax (Watts)	360 ± 31	376 ± 39 ***	364 ± 39	381 ± 46 ***
FCmax (bpm)	188 ± 9	187 ± 12	188 ± 6	186 ± 7
PiAT (Watts)	266 ± 39	275 ± 47	275 ± 41	286 ± 46
FCiAT (bpm)	159 ± 11	154 ± 20*	164 ± 12	159 ± 9*

Figure 3 : Influence de l'entraînement sur les paramètres de puissance (Pmax et PiAT) et de fréquence cardiaque (FCmax et FC iAT) pour les 2 groupes de cyclistes (d'après Mornieux et Stapelfeldt, 2012).

Les prochaines études à ce sujet devront s'attacher à mettre en place une intervention plus longue afin de renforcer l'apprentissage de ce nouveau pattern de pédalage et éventuellement

améliorer la performance. Nous pouvons même supposer qu'un entraînement plus long permettrait des adaptations neuromusculaires. En effet, en adéquation avec la précédente étude, nous pourrions imaginer qu'en compensation de l'augmentation du travail musculaire durant la phase de retour, une réduction de l'activité EMG des muscles gastrocnemius lateralis et vastus medialis durant la phase descendante de la pédale serait observée, permettant une apparition plus tardive de la fatigue en condition de traction sur la pédale.

1.2.2) Contrôle et organisation du mouvement lors de changements de direction

Durant ces travaux postdoctoraux en Allemagne et dans le cadre de prestations de service en analyse du mouvement pour un équipementier, l'intérêt pour une plus grande compréhension des mouvements de un contre un et plus largement des changements de direction s'est fait de plus en plus fort. Ainsi, au travers de ces coopérations avec cet équipementier désirant optimiser le chaussage avec comme objectifs principaux i) l'augmentation du confort, ii) l'amélioration de la performance et iii) la réduction des contraintes articulaires, un changement thématique s'est opéré afin d'étudier le contrôle et l'organisation des mouvements de changement de direction et plus spécifiquement les problématiques de performance et de blessure durant ces mouvements.

Les changements de direction sont une habileté technique, dont les principales caractéristiques sont corrélées à la performance. Ainsi, les caractéristiques de temps d'exécution du mouvement, de temps de prise de décision ou de précision dans l'exécution du geste sont différentes en fonction du niveau d'expertise des joueurs (Figure 4), comme par exemple chez des rugbymen dans l'étude de Gabbett & Benton (2009).

Reactive agility	Elite	Sub-elite
Movement time (s)	2.35 (2.30–2.40)*	2.56 (2.51–2.61)
Decision time (ms)	89.5 (78.2–100.8)*	111.5 (99.0–124.0)
Response accuracy (%)	93.2 (89.5–96.8)*	85.5 (80.6–90.5)

Data are mean (and 95% confidence intervals).
* Denotes significant differences ($p < 0.05$) between groups.

Figure 4: Influence du niveau d'expertise (rugbymen elite vs. sub-elite) sur le niveau de performance lors de l'exécution d'un changement de direction (d'après Gabbett et Benton, 2009).

De plus, les changements de direction sont associés à des contraintes articulaires élevées, pouvant conduire à des blessures, notamment ligamentaire au niveau de l'articulation du

genou (Figure 5), étant donné que ce mouvement engendre au niveau du genou un valgus, combiné à un moment de rotation et une translation du tibia par rapport au fémur, soit l'un des mécanisme classique de rupture du ligament croisé antérieur (Alentorn-Geli et al., 2009).



Figure 5 : Rupture du ligament croisé antérieur du genou lors d'un changement de direction chez un joueur de football (d'après Alentorn-Geli et al., 2009).

Afin d'étudier le contrôle et l'organisation des mouvements de changement de direction et plus spécifiquement les problématiques de performance et de blessure durant ces mouvements, 10 études, intégrant deux travaux de doctorat et ayant conduit à 11 publications, ont été menées. L'ensemble de ces travaux est basé sur une approche biomécanique classique, incluant une mesure cinématique 3D, une mesure des forces de réaction au sol (plateforme de force), ainsi qu'une mesure de l'activation neuromusculaire par électromyographie de surface des muscles pertinents. Les moments articulaires, permettant une analyse des contraintes, sont déterminés par dynamique inverse.

Les changements de direction et plus généralement les mouvements latéraux font partie des déplacements de base dans tous les sports collectifs, voire dans d'autres disciplines comme le tennis. Ce mouvement peut être réalisé exclusivement dans le plan frontal ou de manière plus complexe impliquant les 3 dimensions. L'analyse de ce mouvement et son contrôle neuromusculaire associé sont importants afin de comprendre ses spécificités pour plus tard optimiser la réalisation de cette tâche. Une série de mesures, dans le cadre du doctorat de Mlle Jana Fleischmann, s'est focalisée sur la tâche du saut latéral afin de placer l'athlète dans le plan frontal. Dans une première étude (Fleischmann, Gehring, Mornieux, & Gollhofer, 2010), 15 joueurs de basket ont réalisé cette tâche consistant en la réalisation d'un saut vers la droite avec réception sur la jambe droite avant une poussée la plus explosive possible afin de revenir dans sa position initiale. Quatre séries de sauts étaient réalisées en modulant la distance, à savoir 100% de la distance maximale pouvant être atteinte par le participant, 85% de cette distance maximale, puis 70% et enfin 55% de cette distance maximale. Les principaux

résultats ont montré une augmentation du moment articulaire dans le plan frontal au niveau de la cheville, du genou et de la hanche liée à l'augmentation de la distance de saut latéral. Plus cette distance augmentait et plus les flexions de hanche et de genou ainsi que l'abduction de hanche augmentaient. Cette surcharge induite par la distance de saut latéral était accompagnée d'une augmentation significative de l'activation des muscles de la cuisse et notamment des vasti medialis et lateralis et du semitendinosus. Cependant, l'angle articulaire de la cheville n'était pas influencé par la modulation de la tâche. Une autre stratégie de régulation de la charge lors de ce saut latéral a pu être mise en évidence, à savoir le recours à une rotation externe du pied plus importante (Figure 6).

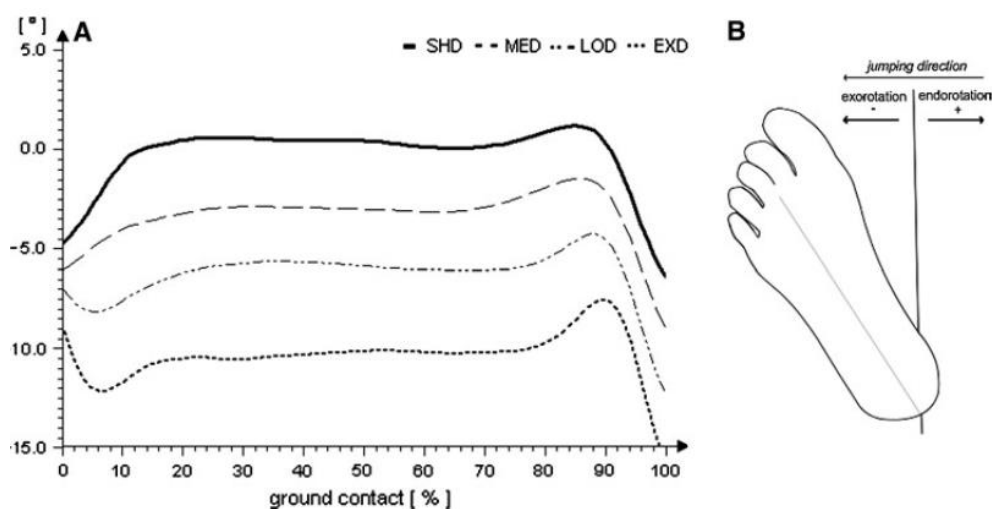


Figure 6 : Influence de la distance du saut latéral (short, medium, long, extreme) sur la rotation externe du pied durant le contact au sol (d'après Fleischmann et al., 2010).

Ainsi, la compensation de la surcharge dans le plan frontal serait tout d'abord opérée par une augmentation de la rotation externe du pied d'appui, permettant peut être de transférer la surcharge frontale dans le plan sagittal étant donné l'augmentation de la flexion du genou et de la hanche. Quoiqu'il en soit, la charge durant un saut latéral apparaît être modulée grâce à des ajustements cinématiques et neuromusculaires au niveau du genou et du bassin et non au niveau de la cheville.

Afin d'explorer plus précisément ces aspects du contrôle du mouvement durant l'exécution d'un mouvement latéral, 10 sujets ont ensuite réalisé une série de sauts latéraux comme décrits précédemment, puis de sauts latéraux sans procéder à la poussée de la jambe d'appui (Fleischmann, Gehring, Mornieux, & Gollhofer, 2011). L'ensemble des paramètres cinétiques, cinématiques et neuromusculaires était à nouveau enregistré lors de ces sauts

réactifs vs. « atterrissages ». Le principal résultat de cette étude démontre à nouveau l'importance de la rotation externe du pied, i) permettant de réguler l'augmentation de la charge consécutive à une distance de saut plus importante (ceci est conforme à l'étude précédente) et ii) apparaissant comme spécifique au saut latéral réactif étant donné que le simple saut sans action de poussée n'engendrait pas de modification de la rotation du pied. Par ailleurs, les forces moyennes verticales et médio-latérales de réaction au sol lors des premiers 150ms de contact étaient supérieures lors de l'exécution d'un saut latéral réactif par rapport à un atterrissage et ne présentait pas de modulation en fonction de la distance de saut (Figure 7). Les moments articulaires s'en trouvaient d'ailleurs fortement augmentés dans la première tâche. Il apparaît ainsi que la contrainte temporelle liée à un temps de contact réduit afin d'exécuter le saut latéral réactif se traduit par une absence de modulation des forces de réaction au sol afin de générer l'impulsion nécessaire. En revanche, lors d'un saut sans poussée successive, le temps de contact plus long permettait au sujet de réduire les forces de réaction au sol.

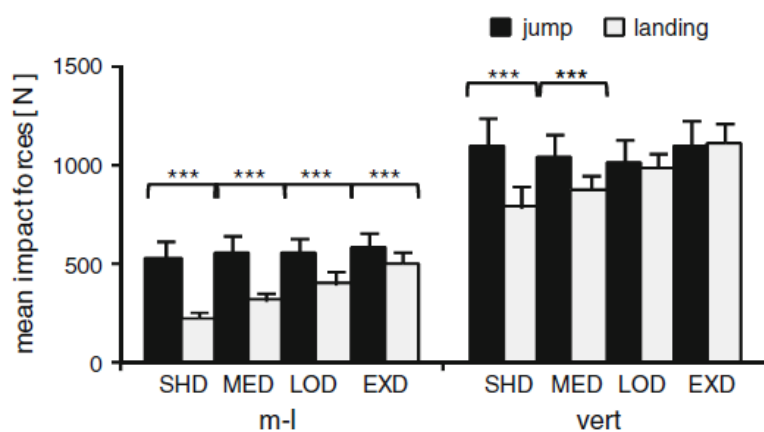


Figure 7: Influence de la distance d'un saut ou d'un atterrissage latéral (short, medium, long, extreme) sur les forces moyennes de réaction au sol (d'après Fleischmann et al., 2011).

En conclusion, dans des sports nécessitant des changements de direction, notamment lorsque ceux-ci ont lieu dans le plan frontal, il apparaît pertinent de mettre en place des exercices à l'entraînement permettant de travailler le placement du pied et plus précisément cette rotation externe afin de prévenir les contraintes articulaires élevées.

Les mouvements latéraux sont souvent réalisés dans un environnement dans lequel l'ensemble des paramètres est difficile à contrôler, voire difficile à prédire, comme par exemple lorsque le mouvement est perturbé par un contact avec un adversaire ou lorsque l'appui au sol est perturbé par une friction chaussure-sol inadéquate (blocage ou glissade). De ce fait, l'analyse des paramètres importants du contrôle du mouvement durant un changement de direction, comme la rotation externe du pied ou le contrôle neuromusculaire de la cuisse

vus précédemment, doit se faire dans un environnement non prévisible voire avec perturbation. Dans ce contexte, une première étude a repris le paradigme du saut latéral réactif durant lequel l'appui au sol était perturbé (Mornieux, Gehring, Tokuno, Gollhofer, & Taube, 2014). A l'instant du contact initial lors de l'exécution de saut latéral réactif, le sol restait soit immobile (simulant un appui stable), ou se déplaçait de 10cm soit vers la droite (simulant une glissade), soit vers la gauche (simulant un blocage) (Figure 8).

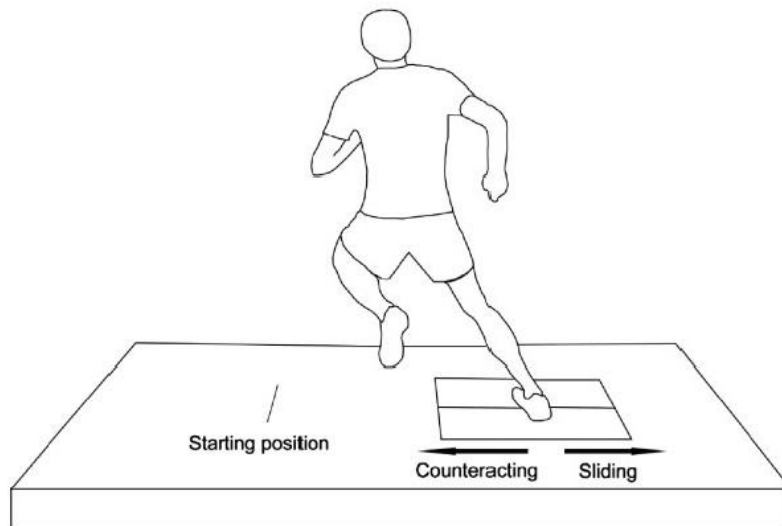


Figure 8 : Représentation des trois modalités d'exécution du saut latéral réactif au moment de l'appui : immobile (stable), glissade vers la droite (sliding) et blocage vers la gauche (counteracting) (d'après Mornieux et al., 2014).

En outre, les 12 sujets ayant participé à cette étude ont réalisé ces trois modalités de saut en ayant ou non au préalable la connaissance de la modalité à venir. Autrement dit, une série de saut était réalisée en indiquant au participant quelle modalité interviendrait, suivi d'une autre série lors de laquelle le sujet ne connaissait pas, parmi les trois modalités, celle qui serait appliquée. Cette tâche permettait de comprendre comment l'individu régulerait son contrôle moteur, et notamment le système de feed-forward utilisé pour ces mouvements rapides, lors de la connaissance d'un nouvel événement perturbant la commande pré-programmée (ici la modalité de saut étant perçue au moment de l'impact, nécessitant de moduler le contrôle moteur mis en place avant l'appui au sol). Cette étude a clairement mis en évidence la nécessité de moduler la cinématique du membre inférieur lors de l'exécution du saut latéral en fonction de la modalité d'exécution (Figure 9 ; predictable setting). Lorsque le participant ne connaissait pas le type de perturbation, l'agencement du membre inférieur au moment du contact initial était identique quelle que soit la modalité, ne lui permettant pas d'ajuster

précisément les paramètres cinématiques de la jambe (Figure 9 ; unpredictable setting). Dans ce cas de figure, la stratégie utilisée par les participants était une augmentation significative de l'abduction de hanche afin de pouvoir répondre à l'ensemble des modalités d'exécution du saut latéral. De plus, les flexions de hanche et de genou étaient significativement réduites dans la condition de glissade lorsque la modalité de saut n'était pas connue, indiquant une moins bonne organisation dans le plan sagittal pour cette condition.

	Predictable setting			Unpredictable setting		
	Stable	Sliding	Counteracting	Stable	Sliding	Counteracting
Hip flexion (°)	29.9 ± 6.5	30.6 ± 7.2	26.1 ± 10.1	26.6 ± 8.2	27.4 ± 7.2 ^a	27.2 ± 7.5
Hip abduction (°)	26.4 ± 7.3	28.4 ± 6.8	28.6 ± 6.4	31.0 ± 6.1 ^a	30.4 ± 5.8 ^a	30.7 ± 6.1 ^a
Knee flexion (°)	23.6 ± 5	26 ± 3.5	23.3 ± 3.8	23.7 ± 4	24 ± 3.7 ^a	23.8 ± 3.7
Knee adduction (°)	-1.1 ± 3.9	-1.5 ± 4.3	-1.3 ± 3.8	-1.7 ± 4.4	-1.6 ± 4.5	-1.3 ± 4.4
Dorsiflexion (°)	-13 ± 8.9	-8.7 ± 8.8	-11.3 ± 8.6	-12.8 ± 6.1	-13.8 ± 8.1	-13.1 ± 7.3
Ankle inversion (°)	12.4 ± 5.5	11.6 ± 5.1	10.8 ± 4.9	10.4 ± 3.6	9.8 ± 4	10.4 ± 4

Figure 9 : Moyennes ± écarts types des angles articulaires du membre inférieur lors de l'exécution d'un saut latéral réactif avec ou sans perturbation (stable vs. sliding vs. counteracting) et avec ou sans la connaissance au préalable de la modalité de saut (predictable vs. unpredictable setting). Ces valeurs ont été évaluées à l'instant du contact au sol (d'après Mornieux et al., 2014).

Ces changements dans l'agencement articulaire de la jambe lorsque la modalité n'était pas connue au préalable engendrait des modifications bien plus importantes encore à 140ms après contact. Ainsi, non seulement l'abduction de hanche était différemment modulée, mais les flexions de hanche, de genou et l'inversion de cheville étaient également différentes de la condition, dans laquelle les participants connaissaient au préalable la perturbation à venir. Cela démontre qu'un contrôle moteur inadéquat durant des mouvements latéraux, notamment dû à un contrôle feed-forward non optimal, engendre une configuration articulaire ne permettant pas de répondre spécifiquement aux différentes situations possibles (glissade, blocage, appui stable) et ayant des conséquences sur la qualité de l'appui au moins pendant 140ms après le contact. Ceci permet de supposer que la performance serait moindre et que les contraintes articulaires pourraient s'en retrouver augmentées. Cette étude introduit l'idée de construire des séances d'entraînement en introduisant de l'incertitude dans l'exécution de la tâche voire de la perturbation.

Si la tâche de saut latéral réactif permet d'isoler le mouvement dans le plan frontal et ainsi d'analyser précisément la dimension médio-latérale du changement de direction, il demeure indispensable de s'intéresser au changement de direction dans son intégralité (cutting

manœuvre). Classiquement dans la littérature, la tâche de changement de direction consiste en la réalisation d'une course linéaire d'approche (souvent entre 4 et 5m/s), suivie d'un changement de direction (30° - 90°). Lors de ce mouvement, le freinage opéré dans les premiers instants de l'appui combiné à un changement de direction, place une contrainte élevée au niveau de l'articulation du genou avec notamment un moment externe d'abduction et de rotation. Comme nous l'avons vu précédemment, l'introduction de l'incertitude dans la réalisation de cette tâche est pertinent et s'opère le plus souvent en proposant, via un signal lumineux, des directions à suivre lors du changement de direction, qui sont présentées de manière aléatoire peu de temps avant l'appui (Figure 10).

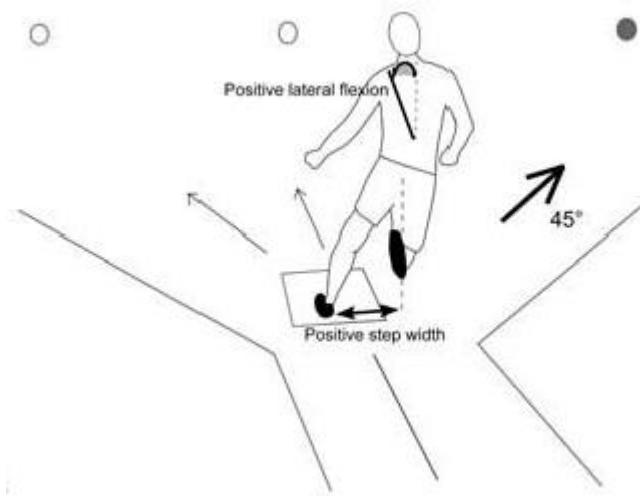


Figure 10 : Protocole expérimental permettant l'étude du changement de direction. Après une course, le participant suit l'indication lumineuse de direction : changement de direction vers la gauche si la lampe de gauche s'allume, tout droit si la lampe du milieu s'allume, changement de direction vers la droite à 45° si la lampe de droite s'allume, comme dans ce cas précis (d'après Mornieux et al., 2014).

Lors d'un changement de direction, la nécessité de réorienter son centre de gravité demande le recours à plusieurs stratégies à différents niveaux que sont l'appui et les différents segments du corps. Le temps de préparation afin d'organiser ce changement de direction, à savoir le temps entre l'information (provenant de la lampe) et l'appui au sol amène une contrainte supplémentaire dans la réalisation de cette tâche. Le but de cette première étude utilisant ce paradigme était d'identifier quelles stratégies étaient défailtantes lorsque le temps de préparation à la tâche était limité (Mornieux, Gehring, Fürst, & Gollhofer, 2014). 13 footballeurs ont réalisé plusieurs changements de direction (à gauche, tout droit, à droite à 45°) selon différentes conditions de préparation du mouvement : soit de manière « anticipée », c'est-à-dire qu'il était annoncé au participant quelle lampe s'allumerait, soit de manière « non-anticipée » et les footballeurs bénéficiaient d'un certain temps de préparation pour prendre l'information et préparer leur mouvement. Dans une première série, ce temps de préparation était de 850ms, puis de 600ms et enfin de 500ms dans les séries suivantes. Les

résultats, à l'instant d'apparition du moment pic d'abduction du genou lors de la phase de contact au sol, soulignent tout d'abord une rotation de la tête dans la nouvelle direction du mouvement, alors que le tronc reste légèrement tourné et incliné à l'opposé. Les paramètres de pose d'appui, à savoir l'écartement de l'appui (step width) et la rotation externe du pied (foot progression angle) décrivaient une pose latérale avec une rotation externe du pied (Figure 11). Ensuite, en prenant la situation « anticipée » comme référence, il est à noter que le caractère « non-anticipée » d'une tâche n'explique pas à lui seul la contrainte supplémentaire dans la réalisation du changement de direction. En effet, dans la condition où les participants disposaient de 850ms entre le signal de changement de direction et la pose du pied au sol, les variables cinématiques étaient comparables à celles de la condition « anticipée ». Ainsi, ne pas connaître à l'avance le type de changement de direction n'altère pas la cinématique de l'athlète à partir du moment où il dispose de suffisamment de temps pour préparer le mouvement (>850ms pour un changement de direction à 45° avec une vitesse d'approche de 5m/s).

	Ant	850 ms	600 ms	500 ms
Head rotation (°)	18.0 ± 11.3	17.6 ± 13.1	13.3 ± 13.5 ^{*a}	13.5 ± 11.5
Trunk rotation (°)	-4.8 ± 11.0	-3.3 ± 11.1	-7.3 ± 10.3 ^{*b}	-8.8 ± 9.9 ^{*b}
Lateral trunk flexion (°)	5.3 ± 6.1	6.1 ± 6.4	10.4 ± 6.4 ^{*a}	13.7 ± 6.9 ^{*a,c}
Step width (m)	0.36 ± 0.03	0.35 ± 0.04	0.36 ± 0.03	0.35 ± 0.03
Foot progression angle (°)	7.8 ± 7.8	9.2 ± 7.9	6.9 ± 9.7	6.3 ± 9.8

Figure 11 : Moyennes ± écarts types des paramètres de réorientation du corps lors du changement de direction à droite à 45°, selon les différentes conditions (anticipée vs. 850ms vs. 600ms vs. 500ms). Ces valeurs ont été évaluées à l'instant de la valeur pic du moment d'abduction du genou lors de la phase de contact au sol. Avec a : différent de Ant et 850ms ; b : différent de 850ms ; c : différent de 600ms (d'après Mornieux et al., 2014).

Ensuite, lors de l'exécution de cette tâche sous contrainte de temps élevée, les participants tournaient moins la tête et le tronc dans la nouvelle direction et augmentaient leur flexion latérale de tronc. Finalement, la stratégie la plus efficace lors d'un changement de direction sous contrainte de temps semble être l'inclinaison latérale du tronc afin d'accélérer médialement le centre de masse. Cette gestion du haut du corps lors des mouvements latéraux est donc une nouvelle variable importante à prendre en compte.

Les mouvements latéraux dépendent de certains ajustements dans le contrôle du mouvement et notamment la rotation du pied lors de l'appui, le contrôle neuromusculaire de la cuisse, l'abduction de hanche, mais également de l'inclinaison du tronc et l'orientation du

tronc et de la tête. L'intégration du haut du corps, et plus précisément son lien avec le pelvis fait référence à la notion de Core Stability (Gollhofer, Mornieux, Weltin, & Gehring, 2015). La capacité à contrôler la position et le mouvement du tronc par rapport au bassin et au membre inférieur définit classiquement le Core Stability et permet une production de force et un transfert de force optimal vers les segments terminaux. C'est le cas par exemple en baseball avec un transfert de force de la jambe d'appui vers le bras lanceur ou en aviron avec un transfert de la barre de pied vers les avirons. La zone anatomique impliquée peut aller des genoux jusqu'aux épaules mais concerne essentiellement le bassin, la région lombo-pelvienne et le tronc. L'activation des différents muscles de cette région, la pression intra-abdominale ainsi que les différentes structures passives du système musculo-squelettique garantissent une stabilité ainsi qu'une production de force de ce maillon dans la chaîne cinétique reliant les extrémités corporelles. Le Core Stability joue donc un rôle essentiel dans la réponse aux perturbations, et notamment celles au niveau du haut du corps, ou bien celles tendant à perturber la chaîne cinétique comme par exemple un appui glissant. L'évaluation du Core Stability se fait classiquement au travers de ses paramètres cinématiques (contrôle de la position et du mouvement) et de ses paramètres neuromusculaires (électromyographie de surface des muscles external obliquus, erector spinae, rectus abdominis). Ce modèle de Core Stability permet d'aborder les questions relatives à la performance lors de tâche sportive mettant en jeu l'ensemble du corps, au travers de cette notion de transfert de force et de stabilité, ainsi que les questions autour de la prévention des blessures, principalement de la charnière lombo-sacrée mais également du genou. Le paradigme du saut latéral réactif avec perturbation de l'appui évoqué précédemment est dans ce cadre particulièrement pertinent.

Ce paradigme a d'ailleurs été utilisé dans une étude de Weltin, Gollhofer, & Mornieux (2016) afin d'étudier les différences de Core Stability entre hommes et femmes. Cette étude est l'une des trois constituant la thèse de doctorat de Mr. Elmar Weltin, co-encadrée avec le Professeur Gollhofer. 12 athlètes féminines et 12 athlètes masculins ont réalisé différents sauts latéraux réactifs (Figure 8) en connaissant au préalable la condition à exécuter. La cinématique 3D du tronc, du bassin, des articulations de hanche et de genou ainsi que la position du pied ont été analysées. Alors que les hommes utilisent une stratégie centrée sur la position du bassin, les femmes adoptent une stratégie basée sur la rotation du tronc (Figure 12). En effet, les hommes démontraient une inclinaison médiale du bassin durant le contact au sol alors que les femmes voyaient leur bassin être inclinée vers l'extérieur. Cette stratégie présente l'avantage d'aligner

le bassin sur le membre inférieur et ainsi permettre un transfert de force lors de la poussée au sol plus efficace.

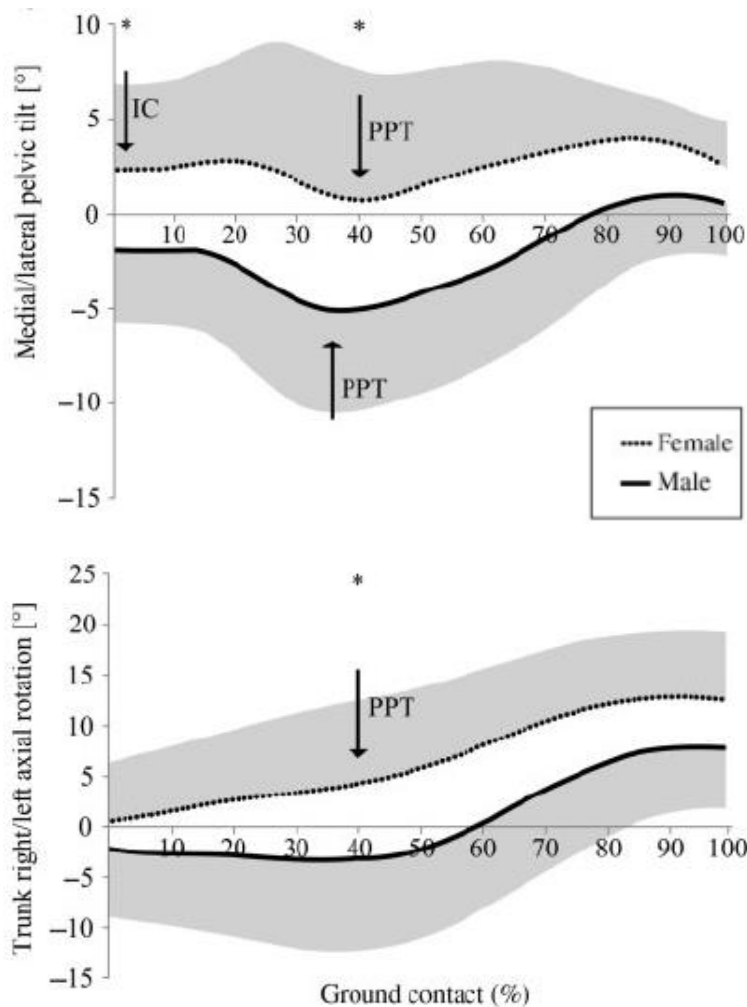


Figure 12 : Evolution de l'inclinaison médiale/latérale du bassin et de la rotation du tronc durant le contact au sol lors d'un saut latéral réactif chez une population de sportifs hommes et femmes (d'après Weltin et al., 2016).

IC : contact initial

PPT : instant de l'inclinaison pic du bassin

Les femmes tournaient le tronc dans la nouvelle direction dès l'impact au sol, alors que les hommes conservaient une rotation du tronc stable durant la phase excentrique avant d'orienter le tronc dans la nouvelle direction lors de la poussée. D'ailleurs, cette instabilité du tronc chez les femmes était associée à une augmentation de la rotation externe de hanche et de l'abduction de genou. Même si les femmes réalisaient la tâche avec succès, il est important de noter que ce type de stratégie menant à une augmentation de la rotation de hanche et d'abduction de genou est lié à une augmentation de la contrainte articulaire au niveau du genou.

Cette première partie a permis de mettre en évidence le rôle important i) de la rotation externe du pied lors de l'appui et ii) du Core Stability dans le contrôle et l'organisation du

mouvement lors de changements de direction. En effet, ces deux aspects du contrôle du mouvement permettent de répondre aux contraintes liées à l'intensité du saut, aux perturbations, à l'incertitude du mouvement, au temps disponible pour organiser le mouvement et sont différentes entre les hommes et les femmes. Ainsi, les protocoles interventionnels visant à optimiser la performance durant les changements de direction devront se baser sur ces éléments.

1.2.3) Optimisation des paramètres biomécaniques de la performance lors de changements de direction

La mise en place de protocoles d'intervention afin d'influencer i) la rotation externe du pied lors de l'appui et ii) le Core Stability, i.e. deux éléments centraux dans le contrôle et l'organisation du mouvement lors de changements de direction, est nécessaire dans le but d'améliorer ou d'optimiser la réalisation de la tâche, voire la performance.

Les précédents travaux de Mlle Jana Fleischmann ont mis en évidence la stratégie d'appui du pied en rotation externe (foot progression angle) lors d'un saut latéral afin de transférer une partie des contraintes articulaires du plan frontal dans le plan sagittal. Cette stratégie s'est avérée efficace lorsque la contrainte était plus forte, soit due à une augmentation de la distance de saut, soit lorsque le temps de contact était réduit afin d'opérer une poussée rapide après contact au sol. Cependant, ce placement en rotation externe du pied tend à tourner l'ensemble du membre inférieur vers l'extérieur et ne semble donc pas fonctionnel d'un point de vue de la performance lors d'une tâche de changement de direction, lors de laquelle l'ensemble du corps devra être réorienté médialement. Ainsi, il serait pertinent de trouver un moyen de limiter cette stratégie de rotation externe du pied et de la compenser autrement. Dans ce cadre, l'intervention sur le chaussage paraît pertinente étant donné son impact direct sur l'appui au sol. L'objectif de ce nouveau chaussage serait de limiter la supination au moment de l'appui afin de réduire l'inversion de la cheville lors des mouvements latéraux et donc la contrainte dans le plan frontal. Ainsi, le sportif pourrait se permettre, pour la même intensité de saut latéral, de réduire la rotation externe du pied lors de l'appui afin d'optimiser la poussée. Différents matériaux peuvent être utilisés afin de moduler la rigidité d'une partie de la semelle permettant un comportement bien spécifique lors de l'appui. De plus, en fonction des caractéristiques de l'appui, il est important de déterminer quelle partie de la semelle doit être manipulée afin de produire le mécanisme recherché. Dans cette optique, un

prototype de chaussure de basketball a été développé pendant plusieurs années (Figure 13). Cette chaussure est équipée d'une structure particulière au niveau médial de l'avant du pied. Cette structure synthétique souple contient quatre cavités permettant une compression à ce niveau et plaçant le pied en pronation. Cet effet apparait ainsi uniquement lors de la pose du pied avec un appui sur l'intérieur et l'avant du pied, c'est-à-dire classiquement lors d'un changement de direction.



Figure 13 : La photo A représente la chaussure de basketball classique. La photo B représente la même chaussure équipée des éléments de compression au niveau médial de l'avant du pied. Ainsi, lors d'un appui sur cette zone (e.g. un changement de direction), la compression médiale de la semelle place le pied en pronation et limite donc l'inversion de la cheville (d'après Fleischmann et al., 2013).

L'étude de Fleischmann, Mornieux, Gehring, & Gollhofer (2013) a testé ce prototype en le comparant à la même chaussure de basketball sans ces éléments compressibles. Pour ce faire, 12 joueurs de basketball ont réalisé des sauts latéraux réactifs avec quatre distances différentes, allant de 100% de la distance maximale de saut à 55% de cette distance. L'utilisation du prototype avec ses éléments compressibles permettait bien d'obtenir l'effet escompté étant donné la réduction significative de l'inversion de cheville observée avec cette chaussure (Figure 14), et ce quelle que soit la distance de saut. En conséquence, l'utilisation de la stratégie de rotation externe du pied afin de diminuer la contrainte articulaire au niveau de la cheville était réduite. Le temps de contact au sol n'était pas significativement modifié.

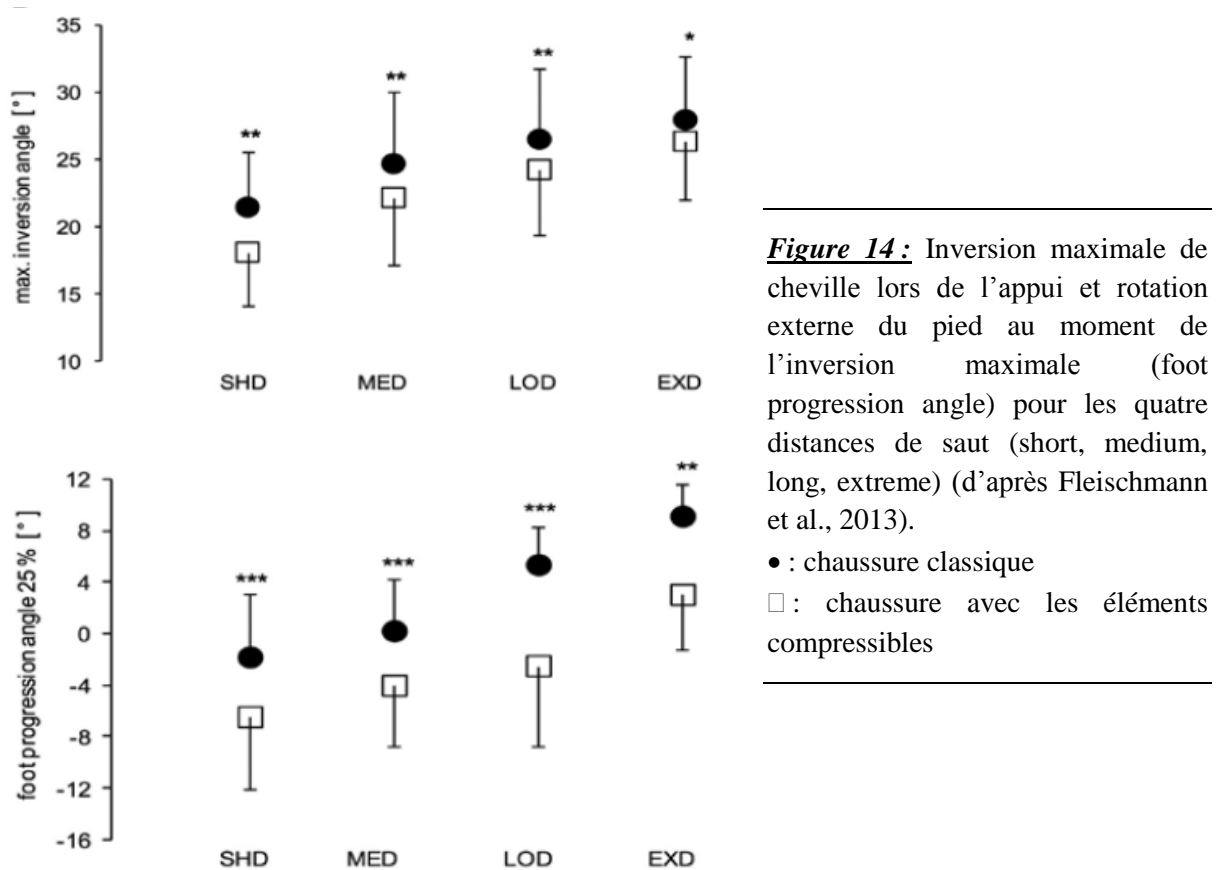


Figure 14: Inversion maximale de cheville lors de l'appui et rotation externe du pied au moment de l'inversion maximale (foot progression angle) pour les quatre distances de saut (short, medium, long, extreme) (d'après Fleischmann et al., 2013).

● : chaussure classique
 □ : chaussure avec les éléments compressibles

Enfin, il est important de noter que ceci avait des conséquences au niveau neuromusculaire. En effet, parmi les muscles de la jambe croisant l'articulation de la cheville ou du genou, l'activité électromyographique des muscles peroneus longus et gastrocnemius lateralis était significativement réduite, exceptée pour la condition de saut extrême. Ainsi, l'utilisation d'un chaussage limitant la supination du pied lors de changement de direction permet de ne pas avoir recours à une stratégie de rotation externe du pied excessive. Cet avantage mécanique était en outre accompagné d'une réduction de l'activation des muscles latéraux de la jambe.

L'importance du Core Stability dans les mouvements de changement de direction étant indéniable, son optimisation semble être une évidence. Une première approche s'est basée sur l'utilisation de matériel en soutien du Core Stability. Les textiles ayant évolués ces dernières années, l'objectif était de concevoir un textile à porter permettant d'aider le Core Stability dans ses fonctions de stabilisation du haut du corps et de transfert des forces vers les segments terminaux. Les habits de compression sont généralement reconnus pour leur effet sur la récupération (chaussettes de compression), mais également plus récemment sur la limitation des amplitudes articulaires au cours de mouvements sportifs. Afin de limiter les amplitudes de mouvement du tronc par rapport au bassin, et notamment lors de perturbations, un habit

devrait recouvrir l'ensemble de la zone fonctionnelle du Core Stability. Une combinaison de compression allant des genoux jusqu'aux épaules a été développée (Figure 15). La compression moyenne était de 5,2 mmHg mais pouvait aller jusqu'à 8,3 mmHg au niveau central du bassin. Au-delà de ses propriétés de compression, des bandes élastiques ont été rajoutées dans le design de la combinaison afin de relier certains points anatomiques permettant un contrôle fonctionnel du tronc par rapport au bassin en 3D.



Figure 15 : Combinaison compressive permettant un support au tronc et au bassin. Les bandes élastiques grises relient le haut du corps aux membres inférieurs au travers du bassin dans la dos, permettant un support fonctionnel du tronc lors de changements de direction (d'après Mornieux et al., 2017).

Dans cette étude (Mornieux, Weltin, Pauls, Rott, & Gollhofer, 2017), une combinaison sur mesure a été confectionnée pour chacune des 12 participantes. Le port de la combinaison engendrait une réduction significative de l'amplitude de mouvement du tronc dans les plans sagittal et transversal lors de mouvements simples et quasi-statiques debout ou allongé sur le dos. Les participantes ont ensuite réalisé les deux protocoles précédemment décrits : les sauts latéraux réactifs avec perturbation et les changements de direction à 45° avec un temps de préparation au mouvement de 650ms. De plus, les participants ont réalisé des tests physiques d'agilité, de détente et de puissance du tronc lors d'un tirage de câble. Globalement, les inclinaisons du tronc dans l'espace n'étaient pas réduites grâce au port de la combinaison durant les mouvements latéraux. Lors du changement de direction à 45°, l'inclinaison latérale était même augmentée (Figure 16). Même si globalement les effets de la combinaison en condition quasi-statique étaient significatifs, la dynamique d'un changement de direction est telle que la combinaison n'apportait pas de soutien au tronc dans le plan frontal. L'effet était même négatif, comme si les participantes réduisaient leur contrôle neuromusculaire du tronc en se reposant sur les effets supposés de la combinaison, entraînant de fait une inclinaison plus importante dans la condition où la combinaison était portée.

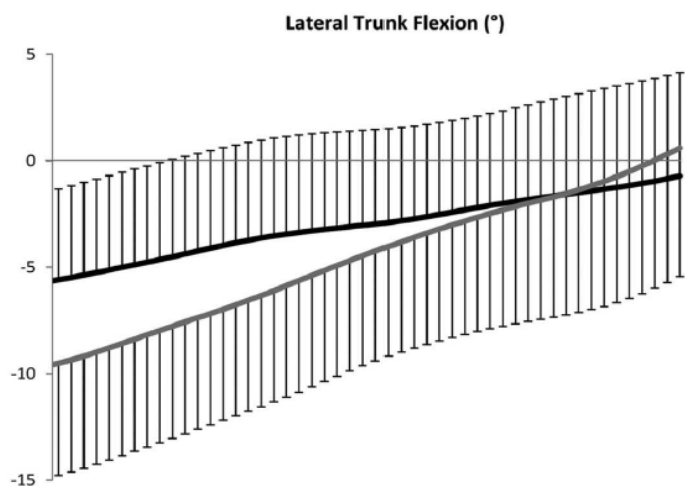


Figure 16 : Evolution de l'inclinaison latérale du tronc au cours de l'appui lors d'un changement de direction à 45° en condition contrôle (trait noir) et lors du port de la combinaison (trait gris). Alors que l'inclinaison est la même en fin d'appui, le port de la combinaison augmente d'environ 5° l'inclinaison latérale du tronc dans la direction opposée en début d'appui (d'après Mornieux et al., 2017).

Les paramètres de performance n'étaient quant à eux pas significativement influencés par le port de la combinaison. Le temps de sprint sur le test d'agilité était le même, la hauteur de saut lors d'un saut avec contre-mouvement était stable à 0,28m et la puissance des rotateurs du tronc était comparable (280W en moyenne).

Si l'utilisation d'un vêtement spécifique au soutien du tronc n'a pas été bénéfique, cela pourrait être du fait de la nature passive de ce support mécanique. Ainsi, replacer l'athlète au cœur d'une étude interventionnelle visant l'amélioration du Core Stability pourrait être plus efficace. Dans ce cadre, Hibbs, Thompson, French, Wrigley, & Spears (2008) soulignent l'importance de l'utilisation de tâches dynamiques et fonctionnelles et de l'augmentation de la résistance (Figure 17).

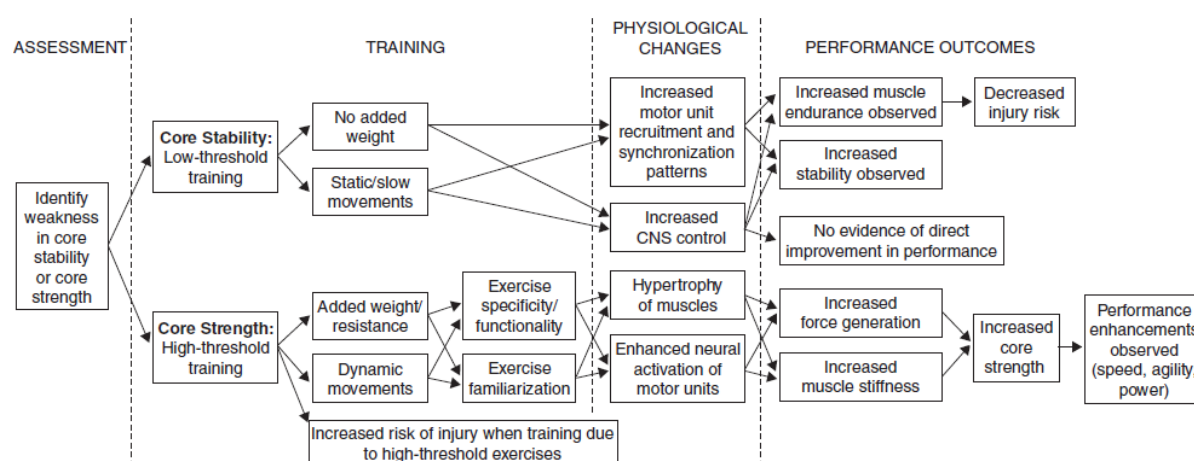


Figure 17 : Bénéfices potentiels d'un entraînement en Core Stability (d'après Hibbs et al., 2008).

Ainsi, un entraînement spécifique visant des adaptations neuromusculaires du Core Stability a été développé (Weltin, Gollhofer, & Mornieux, 2017). Les changements de direction sont des mouvements au cours desquels un faible temps de contact au sol et une vitesse de course élevée en sortie d'appui sont prépondérants pour la performance. De ce fait, de nombreuses études ont utilisé des formes d'entraînement basées sur des exercices plyométriques. Cependant, si un bon Core Stability est censé contrôler et repositionner le tronc par rapport au bassin lors de perturbation, il apparaît pertinent d'inclure cette modalité dans les formes d'entraînement. Ainsi, l'objectif de cette étude était de proposer un programme d'entraînement basé sur des exercices plyométriques classiques en intégrant une perturbation en utilisant le premier paradigme décrit précédemment. 24 athlètes féminines ont participé à l'étude. Un premier groupe de 12 athlètes (groupe contrôle CON) suivait un entraînement plyométrique classiquement décrit dans la littérature basé sur des sauts verticaux, des sauts avec fente avant ou demi-tour et des drop-jumps. Le second groupe de 12 athlètes réalisait un entraînement plyométrique avec perturbation (groupe PPT) à base de sauts latéraux réactifs (Figure 8). Les deux entraînements se déroulaient sur quatre semaines à raison de trois séances de 30 min chaque semaine. La complexité de la tâche et l'intensité de l'exercice étaient progressivement augmentées au fil des séances. L'entraînement plyométrique avec perturbation commençait tout d'abord sans perturbation, puis avec des perturbations linéaires sur 7cm au moment de l'appui en connaissant au préalable la modalité à venir, avant d'augmenter l'amplitude des perturbations (10cm puis 12cm) et de mélanger les différentes modalités sans que l'athlète ne connaisse la condition à venir avant la réalisation de la tâche. Le principal résultat montre une organisation complètement différente du tronc lors d'un saut latéral réactif après entraînement chez le groupe ayant réalisé l'entraînement spécifique (Figure 18). Dès le contact initial, la rotation était significativement moins importante dans la direction opposée au mouvement, pour rapidement évoluer vers la nouvelle direction. Ceci était également vrai pour la rotation du bassin, permettant ainsi une organisation plus pertinente du mouvement vers la nouvelle direction.

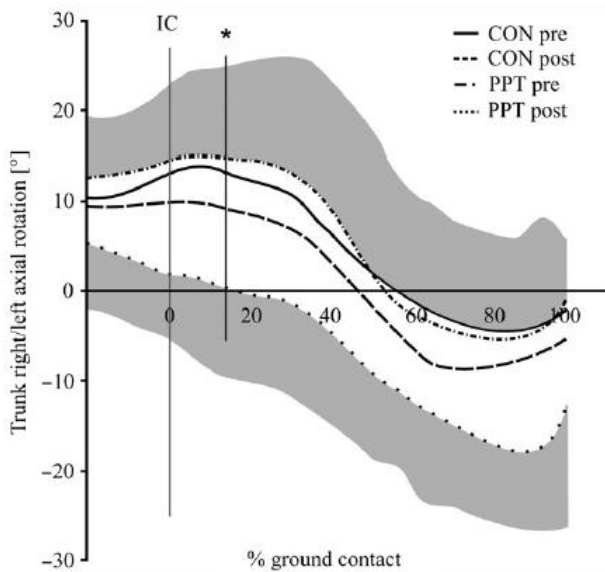


Figure 18 : Evolution de la rotation du tronc au cours de l'appui lors d'un saut latéral réactif chez le groupe contrôle (CON) et le groupe ayant suivi l'entraînement à base de sauts latéraux réactifs avec perturbation (PPT), avant (pre) et après (post) entraînement. Une valeur négative indique une rotation dans la nouvelle direction du mouvement et IC l'instant du contact initial (d'après Weltin et al., 2017).

Cependant, il n'y avait pas de transfert de ces effets lors de la réalisation d'un changement de direction à 45°. La spécificité de la tâche d'entraînement permettait d'influencer le Core Stability durant un saut latéral réactif, mais il semblerait que i) le volume d'entraînement n'ait pas suffi à influencer la réalisation du changement de direction ou bien que ii) la nature plus complexe et plus dynamique du changement de direction ne permette pas cette amélioration du Core Stability. Malgré tout, il semble pertinent d'introduire systématiquement des éléments perturbateurs dans les situations d'entraînement, que ce soit par l'utilisation d'élastiques permettant d'augmenter la résistance ainsi que la perturbation ou des supports instables, notamment au niveau du sol.

Cette deuxième partie a permis de mettre en évidence l'impact du chaussage et d'un vêtement spécifique sur plusieurs facteurs biomécaniques du contrôle du mouvement durant un changement de direction. Même si certaines innovations apportées au chaussage peuvent influencer un aspect bien précis du mouvement, i.e. l'inversion de la cheville et la rotation externe du pied, il semble plus compliqué de modifier un aspect plus global et complexe comme le Core Stability. L'entraînement reste ainsi le meilleur moyen de modifier ces paramètres biomécaniques, surtout s'il est associé à une forme de perturbation. Les entraînements basés sur la technique, comme par exemple un focus sur la pose du pied ou la rotation du tronc sont également efficaces. Un entraînement ciblant plus spécifiquement le Core Stability devrait être envisagé afin de vérifier s'il est possible de modifier le contrôle d'un mouvement aussi complexe et dynamique qu'un changement de direction. La piste actuelle consiste en un entraînement dynamique et fonctionnel étant donné que les programmes quasi-statiques ou peu fonctionnels ont d'ores et déjà montré leurs limites.

1.2.4 Prévention du risque de blessure de la cheville et du genou

Les mouvements latéraux sont également associés à un risque de blessure relativement élevé. Les études épidémiologiques considèrent par exemple que les joueuses de football ont un risque de blessure du ligament croisé antérieur (LCA) de 0,28 pour 1000 pratiques. Si nous restons sur la blessure du LCA, 70% des blessures se produisent en l'absence de contact avec un adversaire ou partenaire, comme par exemple durant un changement de direction. Ce risque de blessure est 4-8 fois plus élevé chez les femmes que les hommes. Dans cette partie, il sera pertinent de reprendre les facteurs biomécaniques les plus importants lors d'un changement de direction, i.e. i) l'appui au sol et plus précisément la rotation externe du pied et ii) le Core Stability, et de comprendre leur influence sur le risque de blessure lors de ces mouvements. Le contrôle de l'inclinaison latérale du tronc est notamment un élément clé dans la prévention des blessures selon le modèle mécanique de Hewett & Myer (2011), décrit sur la figure suivante (Figure 19).

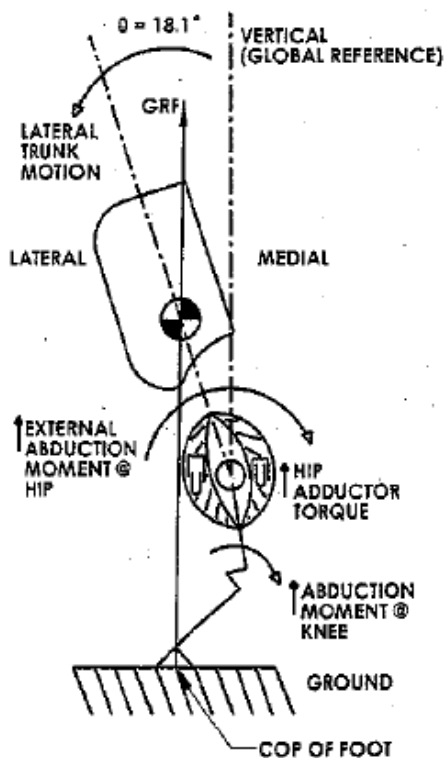


Figure 19 : Modèle mécanique reliant 1) les mouvements latéraux du tronc, 2) le moment d'abduction de hanche et 3) le moment d'abduction du genou, i.e. la contrainte principale expliquant les blessures ligamentaires de cette articulation (d'après Hewett et Myer, 2011).

Lors de tests de chaussure de football utilisant le paradigme du changement de direction, un participant a réalisé plusieurs essais avant de se blesser au niveau de l'articulation de la cheville (Gehring, Wissler, Mornieux, & Gollhofer, 2013). L'objectif de cette étude était ainsi de décrire l'ensemble des paramètres cinétiques, cinématiques et neuromusculaires de la jambe lors d'une entorse de cheville (traumatisme en supination), caractérisée par deux jours

de douleur et de gonflement de l'articulation et un arrêt de pratique sportive pendant 7-10 jours. Durant ces mesures, le paradigme de base (Figure 10) était légèrement modifié, puisqu'une condition de demi-tour remplaçait la course en ligne droite dans le cas où la lampe du milieu s'allumait. 16 demi-tours ont été réalisés avec succès avant la blessure et ont permis de constituer « une zone de confort » des différents paramètres biomécaniques à laquelle pouvait être comparé le demi-tour ayant conduit à la blessure. Le paramètre cinématique le plus évident à observer lors de ce type d'entorse est sans doute l'inversion de cheville (Figure 20). Cependant, on s'aperçoit que ce dernier ne démontrait pas le plus de différence avec les essais réalisés avec succès, même si l'inversion était belle et bien supérieure et notamment lors d'un second mouvement aux 160ms après contact au sol. La flexion plantaire et la rotation interne du pied jouent elles aussi un rôle prépondérant dans le mécanisme de blessure.

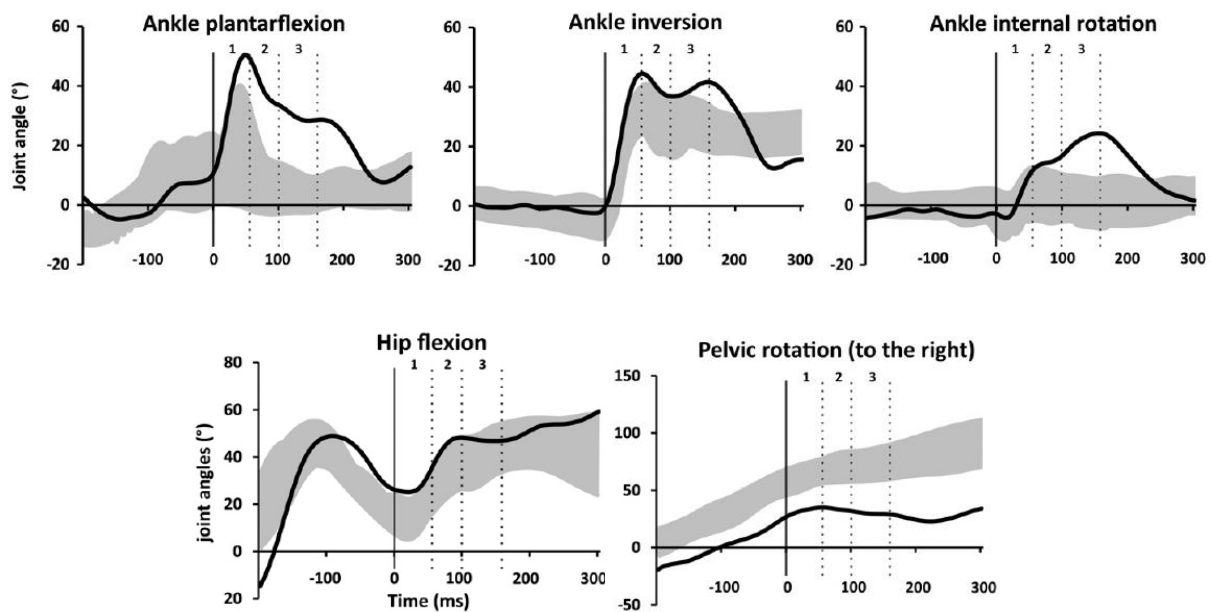


Figure 20 : Paramètres cinématiques du membre inférieur lors du demi-tour ayant engendré la blessure (trait noir) en comparaison de l'ensemble des essais réalisés avec succès (zone grise). La description porte sur les 200ms avant le contact au sol et les premiers 300ms après le contact initial. Les zones 1, 2 et 3 sont basées sur les points caractéristiques de l'inversion de cheville (d'après Gehring et al., 2013).

A un niveau anatomique supérieur, il est intéressant de remarquer que la flexion de hanche et la rotation de bassin sortaient de la norme bien avant le contact au sol (Figure 20), démontrant toute l'importance de la phase de préparation d'un mouvement dans la prévention de blessure. Plus précisément, à 200ms avant le contact au sol, le sujet avait alors une ouverture de hanche et une rotation du bassin vers la gauche alors que le mouvement requérait une flexion de hanche et une rotation du bassin vers la droite. Les moments articulaires de la cheville s'en

trouvaient également altérés, notamment de par la forte augmentation du moment d'inversion dans la zone 3 définie sur la figure 20. Ces altérations cinématiques nous démontrent bien l'importance du contrôle neuromusculaire des fléchisseurs dorsaux et éverseurs de cheville afin de compenser ou limiter le traumatisme en supination. Même si les signaux EMG ne montraient pas de différences aussi nettes que les paramètres cinématiques, on pouvait noter une tendance à la diminution de l'activation des muscles tibialis anterior et peroneus longus immédiatement après le contact au sol (Figure 21). Cette baisse d'activation neuromusculaire était suivie d'une sur-activation dans la zone 3 afin de limiter le traumatisme.

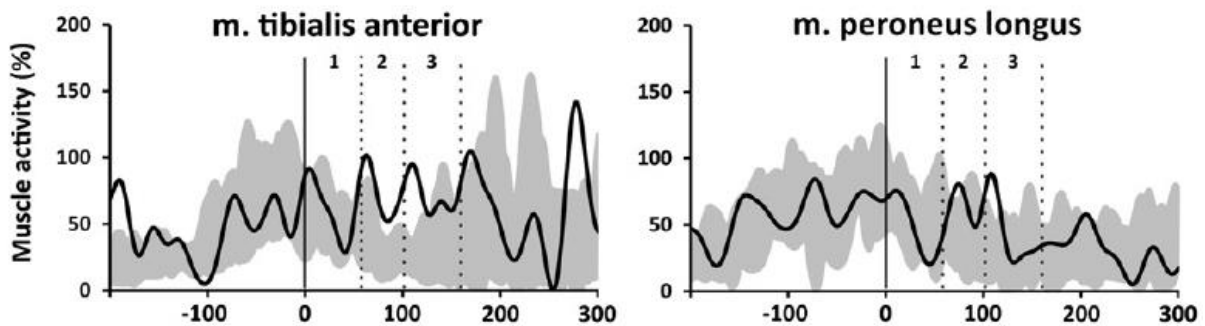


Figure 21 : Paramètres neuromusculaires de la jambe lors du demi-tour ayant engendré la blessure (trait noir) en comparaison de l'ensemble des essais réalisés avec succès (zone grise). La description porte sur les 200ms avant le contact au sol et les premiers 300ms après le contact initial. Les zones 1, 2 et 3 sont basées sur les points caractéristiques de l'inversion de cheville (d'après Gehring et al., 2013).

Ainsi, les paramètres cinématiques du Core Stability, la phase préparatoire au mouvement et la contribution neuromusculaire ne sont pas à négliger lors d'études s'intéressant aux mécanismes d'entorse de cheville.

Si le Core Stability joue un rôle non négligeable dans la prévention des blessures de cheville, son impact sur celles du genou serait encore plus important (Figure 19). Si ce lien théorique entre le contrôle du tronc dans le plan frontal et les contraintes articulaires du genou repose sur un modèle, une étude récente (Weltin, Mornieux, & Gollhofer, 2015) a pu le vérifier. 24 participants (12 femmes et 12 hommes) ont réalisé des sauts latéraux réactifs afin de placer la composante principale du mouvement dans le plan frontal. Durant ces sauts, l'inclinaison latérale du tronc, ainsi que les moments d'abduction de hanche et de genou étaient déterminés. Cette étude confirme l'existence d'une association globale entre l'inclinaison latérale du tronc et le moment d'abduction du genou ($r = 0,67$). Autrement dit,

plus l'inclinaison latérale du tronc augmente et plus la contrainte articulaire au niveau du genou est importante (Figure 22).

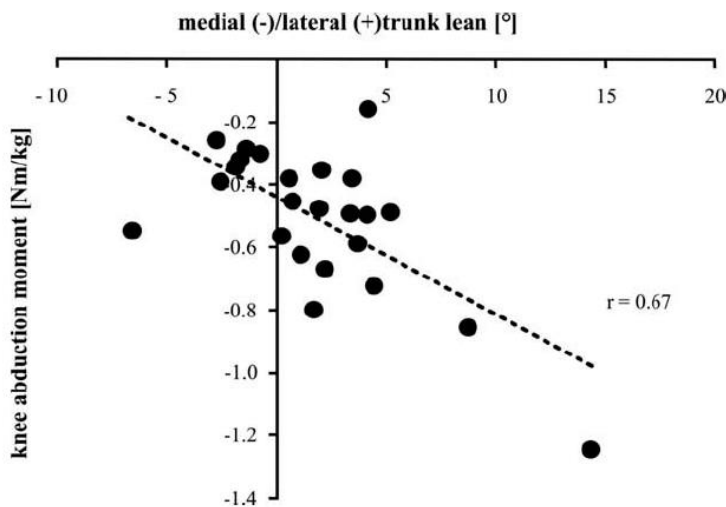


Figure 22 : Relation entre l'inclinaison latérale du tronc et le moment d'abduction du genou lors d'un saut latéral réactif (d'après Weltin et al., 2015).

Si l'on regarde plus en détail les relations entre le haut du corps et le membre inférieur, on remarque une association entre l'inclinaison latérale du tronc et le moment d'abduction de hanche ($r = 0,55$), ainsi qu'une relation entre le moment d'abduction de hanche et le moment d'abduction du genou ($r = 0,78$). Cependant, il n'y avait pas de différence dans ces corrélations entre femmes et hommes. Les paramètres neuromusculaires ont quant à eux montré une influence du genre. Tout d'abord, durant la phase de pré-activation, le *gluteus medius*, permettant l'abduction de hanche, était activé plus tardivement chez les femmes. Ceci pourrait souligner une phase de préparation au saut latéral réactif moins efficace chez les femmes. Ensuite, l'intensité d'activation du *biceps femoris* était plus élevée chez les femmes que les hommes durant cette phase de pré-activation, démontrant une nécessité chez les femmes d'une plus grande activation des ischio-jambiers permettant une protection du ligament croisé antérieur.

La même question s'est posée lors de l'exécution d'un mouvement latéral plus complexe, i.e. un changement de direction à 45° . Dans l'étude décrite précédemment (Mornieux et al., 2014), l'association entre l'inclinaison latérale du tronc et le moment d'abduction du genou a été évaluée (Figure 23). Même si la relation était relativement faible ($r = 0,41$), l'inclinaison du tronc dans le plan frontal explique partiellement les variations de moment d'abduction du genou lors d'un changement de direction et doit retenir l'attention des entraîneurs lors de la mise en place des séances d'entraînement afin d'améliorer le Core Stability.

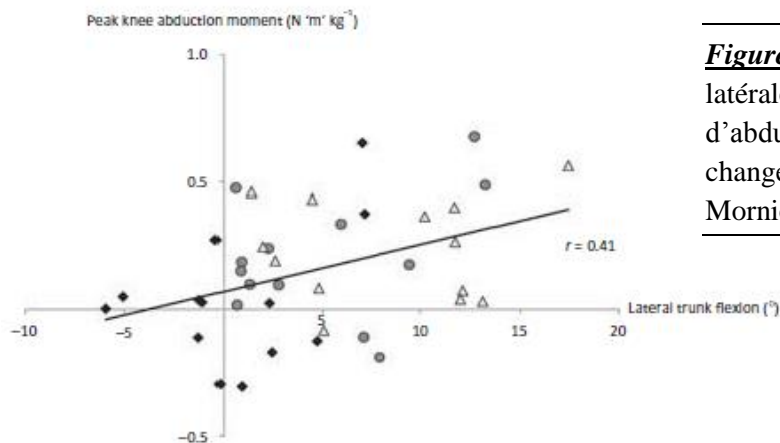


Figure 23 : Relation entre l’inclinaison latérale du tronc et le moment d’abduction du genou lors d’un changement de direction à 45° (d’après Mornieux et al., 2014).

Ainsi, il apparaît pertinent d’améliorer le contrôle du haut du corps également à des fins de prévention de blessure. Les programmes d’entraînement en Core Stability évoqués précédemment auraient alors un objectif double i) d’optimisation de la performance et ii) de prévention de blessures, et notamment au niveau du genou.

II. ACTIVITES D'ENSEIGNEMENT ET ADMINISTRATIVES

II.1) Enseignements

- Maître de Conférences

Université de Lorraine, Faculté des Sciences du Sport (poste n°74-MCF-0386) **2013-...**

- Enseignant Chercheur

Université Albert-Ludwigs de Freiburg (All), Institut du Sport et des Sciences du Sport
2009-2013

- Attaché Temporaire d'Enseignant et de Recherche

Université Jean Monnet de Saint Etienne, Département STAPS **2003-2005**

- Enseignant vacataire

Université Jean Monnet de Saint Etienne, Département STAPS **2001-2003**

Cours	Niveau	Type	Volume annuel	Universités	Années
Analyse du mouvement	Licence	CM (TD)	16h (104h)	Univ. Lorraine	2013-...
	Licence	TD	44h	Univ. Freiburg	2009-13
	Licence	CM (TD)	26h (24h)	Univ. St Etienne	2003-05
	Licence	TD	42h	Univ. St Etienne	2001-03
Analyse du mouvement et entraînement sportif	Licence	CM (TD)	6h (12h)	Univ. Lorraine	2013-...
	Licence	TD	28h	Univ. Freiburg	2002-13
	Licence	TD	8h	Univ. St Etienne	2002-04
Biomécanique	Master	CM (TD)	25h (6h)	Univ. Lorraine	2013-...
	Master	TD	11h	Univ. Freiburg	2009-13
	Master	CM (TD)	8h (8h)	Univ. St Etienne	2003-05
Méthodologie et traitement du signal	Master	CM (TD)	12h (6h)	Univ. Lorraine	2013-...
Outils de mesure	Master	TD (TP)	20h (18h)	Univ. Lorraine	2013-...
APS Football	Licence	TD (TP)	10h (42h)	Univ. St Etienne	2001-05

II.2) Administratif

Responsabilités pédagogiques:

- Responsable pédagogique du Master1 STAPS IEAP parcours-type MPS, Faculté des Sciences du Sport à l'Université de Lorraine (2018-...).
- Responsable pédagogique du Master2 STAPS parcours-type EPAMP, Faculté des Sciences du Sport à l'Université de Lorraine (2014-2018).
- Membre de la commission pédagogique et du jury du Master STAPS, Université de Lorraine (2014-...).

Responsabilités administratives:

- Membre du Conseil de Gestion de la Faculté des Sciences du Sport de Nancy (2017-...).
- Membre du Conseil d'Administration de l'Université Jean Monnet de Saint Etienne, assesseur étudiant (2002-2003).
- Membre du Conseil de Discipline de l'Université Jean Monnet de Saint Etienne (2002-2003).
- Membre du Conseil de la Documentation de l'Université Jean Monnet de Saint Etienne (2002-2003).

Responsabilité en matière de programme de recherche:

- Responsable du laboratoire de biomécanique d'évaluation des athlètes de haut-niveau à Freiburg (2007-2013)

Autres:

- Membre de l'Association Stéphanoise des Etudiants Chercheurs (2001-2003).

III. ENCADREMENTS ET JURYS

III.1) Master 2 soutenu

- M. Schwartz, Master 2 Recherche Evaluation de la Performance et des Adaptations Motrices et Physiologiques (Université de Lorraine, 2018)
- R. Saricos, Master 2 Recherche Evaluation de la Performance et des Adaptations Motrices et Physiologiques (Université de Lorraine, 2017)
- M. Godfrin, Master 2 Recherche Evaluation de la Performance et des Adaptations Motrices et Physiologiques (Université de Lorraine, 2017)
- M. Mastouri, Master 2 Recherche Evaluation de la Performance et des Adaptations Motrices et Physiologiques (Université de Lorraine, 2016)
- N. Balthazard, Master 2 Recherche Evaluation de la Performance et des Adaptations Motrices et Physiologiques (Université de Lorraine, 2016)
- Q. Lallemand, Master 2 Recherche Evaluation de la Performance et des Adaptations Motrices et Physiologiques (Université de Lorraine, 2015)
- M. Pauls, Master 2 Recherche Mouvement et Santé (Université de Freiburg, Allemagne, 2014)
Publication dans le cadre de ce travail de Master : Mornieux, G., Weltin, E., Pauls, M., Rott, F., & Gollhofer, A. (2017). Influence of a full body compression suit on trunk positioning and knee joint mechanics during lateral movements. Journal of Applied Biomechanics, 33, 261-267.
- B. Glunz, Master 2 Recherche STAPS (Université de Freiburg, Allemagne, 2012)
- P. Fürst, Master 2 Recherche STAPS (Université de Freiburg, Allemagne, 2012)
Publication dans le cadre de ce travail de Master : Mornieux, G., Gehring, D., Fürst, P., & Gollhofer, A. (2014). Anticipatory postural adjustments during cutting manoeuvres in football and their consequences for knee injury risk. Journal of Sports Sciences, 32, 1255–1262.
- S. Wissler, Master 2 Recherche STAPS (Université de Freiburg, Allemagne, 2010)
- O. Kurz, Master 2 Recherche STAPS (Université de Freiburg, Allemagne, 2010)
- R. Schween, Master 2 Recherche STAPS (Université de Freiburg, Allemagne, 2009)

III.2) Thèse d'université

- Thèse de Mr. Y. Duchene, Université de Lorraine, EA₃₄₅₀ DevAH. Co-direction (ACT du 11/12/2017) avec le Pr. G. Gauchard. Place du Core Stability dans les interactions posture-mouvement et influence sur la performance et le risque de blessure. **2017-...**
- Thèse de Mr. E. Weltin, Institut für Sport und Sportwissenschaft, Universität Freiburg (All). Co-direction avec le Pr. A. Gollhofer. Knee joint control during lateral movements: influence of gender, trunk control and training. **2010-2015**

Publications dans le cadre de ce travail de Thèse :

Weltin, E., Gollhofer, A., & Mornieux, G. (2017). *Effects of perturbation or plyometric training on core control and knee joint loading in women during lateral movements. Scandinavian Journal of Medicine & Science in Sports, 27, 299–308.*

Weltin, E., Gollhofer, A., & Mornieux, G. (2016). *Effect of gender on trunk and pelvis control during lateral movements with perturbed landing. European Journal of Sport Science, 16, 182–189.*

Weltin, E., Mornieux, G., & Gollhofer, A. (2015). *Influence of gender on trunk and lower limb biomechanics during lateral movements. Research in Sports Medicine, 23, 265–277.*

Thèse d'exercice :

Thèse de Pharmacie de Mr. F. Dubois, Université de Lorraine. Co-direction avec la Pr. B. Faivre. Intérêt de la compression dans l'amélioration et la préservation du retour veineux chez le sportif pratiquant la course à pied. **2018-...**

III.3) Jury

- Thèse de R. Bini, PhD Student, Auckland University of Technology (NZ). Effects of saddle position on pedalling technique and methods to assess pedalling and kinematics of cyclists and triathletes (2011). Rapporteur externe.
- Membre de la commission pédagogique et du jury du Master STAPS, Université de Lorraine (depuis 2014).
- Commission de recrutement : Poste de MCF (section 74, CNU), Université de la Lyon I (UCBL), Mai 2016.

IV. PROJET SCIENTIFIQUE

IV.1) A court terme : Place du Core Stability dans les interactions posture-mouvement et influence sur la performance et le risque de blessure

Ces travaux seront menés dans le cadre de la Thèse de Mr. Duchene (Contrat Doctoral, ED BIOSE, Université de Lorraine, 2017-2020) et directement dans la continuité de ces travaux d'HDR. Ce projet a débuté il y a peu et permettra dans les deux ans à venir de continuer à analyser la place du Core Stability dans l'optimisation de la performance et la prévention de blessures, et notamment au niveau du genou, mais en ajoutant une approche sur le contrôle postural. Ce projet se positionne prioritairement dans l'axe de recherche « Exercice, Entraînement, Performance » de l'EA 3450 DevAH « Développement, Adaptation, Handicap » (directeur : Pr Philippe Perrin) pour la contractualisation 2018-2022.

L'équilibre postural de la station debout est régulé sur la base d'un tonus postural de fond, dépendant de réflexes posturaux émanant des systèmes visuels, vestibulaires et somatokinesthésiques et d'un programme postural qui implique des niveaux de contrôle moteur très élevés (Massion & Woollacott, 1996). Le contrôle postural doit souvent faire face à des situations complexes, dans lesquelles le sujet ne doit pas simplement contrôler des oscillations de base. Ces situations, rencontrées dans la vie courante, nécessitent une habileté à gérer les situations de conflit sensoriel ou à répondre aux perturbations externes ou internes en termes de direction, d'amplitude et de vitesse. Etant le segment le plus inertiel du corps humain, le tronc est un module fondamental dans le contrôle de la posture, notamment dans les situations dynamiques, ce qui renvoie à nouveau à l'importance du Core Stability dans la motricité humaine. Ainsi, il serait primordial de déterminer si l'optimisation du Core Stability favorise le contrôle postural. De plus, différents travaux présentés dans cette HDR suggèrent une influence du Core Stability sur la performance sportive et le risque de blessure, notamment au niveau du genou. Cependant, le nombre d'études ayant démontré qu'une amélioration du Core Stability était corrélée à une augmentation de la performance reste faible, si ce n'est en natation (Weston, Hibbs, Thompson, & Spears, 2015) ou en baseball (Chaudhari, McKenzie, Borchers, & Best, 2011). Par ailleurs, si les paramètres cinématiques liés au Core Stability, que sont l'orientation en trois dimensions du thorax et du pelvis, sont connus, les mécanismes neuromusculaires associés à un meilleur Core Stability ne sont pas suffisamment étudiés et évalués dans la littérature. Ainsi, ce projet vise à déterminer la place du Core Stability dans le contrôle moteur, et plus précisément sur les interactions posture-

mouvement, depuis ses bases mécanistiques jusqu'à son impact sur la performance et le risque de blessure.

Afin de répondre à cet objectif, trois groupes de sportifs ont été sélectionnés sur la base des contraintes posturales liées à leur discipline. Du fait de fortes contraintes posturales en raison de la spécificité gestuelle, trente karatekas, licenciés à la Fédération Française de Karaté et Disciplines Associées ou à la Fédération Luxembourgeoise des Arts Martiaux, seront inclus. Trente athlètes de triple saut et de 110 mètres haies, licenciés à la Fédération Française d'Athlétisme, seront également inclus en raison de spécificités posturales liées aux reprises d'appui instables dans le plan frontal et sagittal. Enfin, trente handballeurs, licenciés à la Fédération Française de Handball, seront inclus en raison de spécificités posturales liées aux différents changements de direction. De plus, les contraintes articulaires au niveau du genou seront différentes dans ces trois groupes, i.e. les plus faibles chez les karatekas et les plus élevées chez les handballeurs.

La première phase visera à comprendre les relations entre Core Stability et contrôle postural (CS-CP) dans le cadre de situations posturales complexes (conflit sensoriel, tâches dynamiques), telles que celles rencontrées dans certaines situations de la vie quotidienne. Les tâches posturales en conflit sensoriel seront réalisées lors du test d'organisation sensorielle, réalisé dans une cabine de posturographie dynamisée de type EquiTest® (Neurocom, Clackamas, OR, USA), qui évalue les capacités du sujet à utiliser séparément les entrées visuelles, vestibulaires et somatosensorielles. Un score d'équilibre est calculé en comparant les oscillations antéropostérieures du sujet durant chaque essai de 20 secondes (Gauchard, Parietti-Winkler, Lion, Simon, & Perrin, 2013). Les tâches posturales lors de perturbations exogènes seront effectuées sur la base de perturbations inattendues en translation du support. Pour chaque tâche, les paramètres du contrôle postural basés sur le centre de pression (trajet, surface, vitesse) et reflétant les stratégies de cheville et de hanche (angles articulaires) seront déterminés. Le Core Stability sera quant à lui évalué à la fois dans ses composantes biomécaniques cinématiques et neuromusculaires. L'analyse cinématique sera effectuée à l'aide de deux capteurs inertiels (IMU, Captiv, France) placés au niveau du torse et du bassin. Les paramètres cinématiques d'inclinaison du thorax et du pelvis dans les trois plans pourront alors être relevés. L'analyse électromyographique sera effectuée à l'aide de 10 capteurs EMG (Delsys, Natick, MA, USA) placés au niveau des muscles rectus abdominis, obliquus externus abdominis et erector spinae pour le tronc, ainsi que les muscles gluteus medius et gluteus maximus pour le bassin. L'analyse des régressions linéaires entre les paramètres du contrôle postural et ceux du Core Stability (cinématiques et neuromusculaires) durant les deux

situations posturales complexes permettra de vérifier l'hypothèse principale d'une relation existante entre CS-CP pour nos trois populations d'athlètes.

La deuxième phase visera à déterminer le Core Stability chez les différentes populations d'athlètes dans le cadre d'une tâche motrice simple mais exigeante sur le plan du contrôle du thorax et du pelvis, tel un changement de direction avec incertitude directionnelle (Figure 10). Les paramètres cinématiques d'inclinaison du thorax, du pelvis et d'angle articulaire du genou dans les trois plans (SIMI, Unterschleissheim, Allemagne) lors du contact au sol durant les différents sauts ainsi que les forces de réactions au sol en 3 dimensions (Bertec, Columbus, USA) seront mesurés ainsi que les composantes électromyographiques du Core Stability décrites lors de la phase 1. Ainsi, les paramètres de contrôle du Core Stability seront analysés, de même que les moments articulaires au niveau du genou par dynamique inverse, afin de déterminer la contrainte articulaire lors des différents essais. Les hypothèses sont i) que les paramètres cinématiques et neuromusculaires du Core Stability sont les meilleurs chez les handballeurs mais également supérieurs chez les athlètes par rapport aux karatékas, et ii) que les moments articulaires du genou sont les plus faibles chez les handballeurs mais également inférieurs chez les athlètes par rapport aux karatékas.

Enfin, la troisième phase visera à déterminer le niveau de Core Stability pour les athlètes lors d'une tâche spécifique et emblématique de leur discipline sportive et à suivre ce niveau de manière longitudinale sur une saison pour le mettre en relation avec leur niveau de performance et le risque de blessure. Les expérimentations auront lieu sur le terrain d'entraînement des différents athlètes. Cette évaluation fonctionnelle focalisera sur un geste technique emblématique de chaque activité, à savoir un Ura Mawashi Geri jambe avant pour les karatekas (coup de pied circulaire de l'intérieur vers l'extérieur), un saut en cloche pour les athlètes et un changement de direction rapide pour les handballeurs. Le niveau de Core Stability (fort ou faible) ainsi que la performance lors de la réalisation de chaque gestuelle spécifique seront suivis durant une saison sportive. Pour ce faire, les paramètres cinématiques et électromyographiques du Core Stability (en utilisant le matériel et la méthodologie de mesure décrits lors de la phase 1) seront évalués au minimum lors de la phase de pré-saison, à la fin de la phase de préparation et en fin de saison, et idéalement toutes les six à huit semaines pendant un an. La performance lors de l'exécution des gestes techniques spécifiques sera évaluée à l'aide de paramètres cinématiques simples (analyse vidéo), tels la précision du coup de pied et la vitesse d'exécution lors des phases aller et retour pour le Ura Mawashi Geri, le temps de contact et la longueur de saut pour la cloche et le temps de contact et la vitesse de course pour le changement direction. Par ailleurs, un suivi des performances et des

blessures au niveau du genou et de la charnière lombo-sacrée sera également réalisé. L'hypothèse principale est que l'entraînement améliore i) le Core Stability et ii) la performance pour nos trois populations d'athlètes. De plus, le niveau de Core Stability sera mis en relation avec le rapport de suivi longitudinal des blessures.

L'originalité de ce projet est donc d'étudier sur trois populations sportives différentes la place du Core Stability dans le contrôle de l'équilibre, en particulier dans des situations posturales complexes liées au conflit sensoriel ou à la réalisation d'une tâche dynamique, afin d'alimenter le cadre théorique dans le domaine du contrôle moteur (phase 1). L'importance du Core Stability pourra ensuite être déterminée lors de la réalisation d'un acte moteur i) simple dans le cadre contrôlé de l'expérimentation en laboratoire lors de la phase transversale (phase 2), puis ii) complexe lié à une discipline sportive dans le cadre d'un suivi longitudinal de terrain (phase 3), afin d'évaluer son rôle sur la performance et le risque de blessure. Si l'ensemble des hypothèses est vérifié, il sera prouvé que le Core Stability est un facteur important du contrôle postural et de la performance motrice (interactions posture-mouvement) et de manière plus pratique de la performance sportive, notamment dans la qualité de la réalisation du geste et la limitation des contraintes articulaires (genou). A terme, les perspectives de ce projet seraient de fournir des solutions pour les athlètes, entraîneurs et développeurs de matériel sportifs, que ce soient dans les domaines des programmes d'entraînement et/ou des textiles sportifs, afin d'améliorer la performance et prévenir le risque de blessure.

IV.2) A moyen terme : Place du Core Stability en aviron

Ce projet s'inscrit dans le cadre d'une collaboration avec le Pôle France Aviron à Nancy ayant conduit à deux travaux de Master ces trois dernières années. L'objectif principal est l'évaluation du Core Stability en aviron afin d'optimiser la préparation physique et l'aménagement de l'embarcation en fonction des caractéristiques morphologiques et techniques des rameurs. Afin de lancer ce projet, un premier financement a été obtenu : Enveloppe « Trajectoire » 2018. Appels à projet BMS (G2-AAP-BMS_003_186_INCITATIF). Ce projet se positionne prioritairement dans l'axe de recherche « Exercice, Entraînement, Performance » de l'EA 3450 DevAH « Développement, Adaptation, Handicap » (directeur : Pr Philippe Perrin) pour la contractualisation 2018-2022.

Il existe deux phases principales dans le coup d'aviron : la propulsion et le retour. Lors de la propulsion, les rames sont en appui dans l'eau, le rameur tend ses jambes et ramène ses bras vers le tronc qui a aussi une action extensive pour déplacer la coque sur l'eau. Le retour ne permet pas de propulser le bateau car les rames sont hors de l'eau. L'ensemble bassin-tronc doit donc transmettre de manière optimale la force provenant des membres inférieurs vers les membres supérieurs, d'où l'importance d'une parfaite transmission des forces dans cette zone charnière qu'est le bassin, tant au niveau des composantes de puissance que des composantes d'équilibre (Baudouin & Hawkins, 2002). Alors que plusieurs études dans la littérature ont mesuré les forces au niveau de la barre de pied et des dames de nage afin de caractériser la performance en aviron d'un point de vue de la gestuelle (Colloud, Bahuaud, Doriot, Champely, & Chèze, 2006), ou en se focalisant sur l'asymétrie dans le coup d'aviron (Warmenhoven, Smith, Draper, Harrison, Bargary, & Copley, 2018), il est dommage que ces analyses mécaniques n'aient pas pris en compte le Core Stability de l'athlète, malgré l'importance du transfert de force et une contrainte de stabilité évidente lors des phases de propulsion et de retour lors du coup d'aviron. A ce titre, un travail récent de Master dans notre équipe a permis de démontrer la faisabilité de l'évaluation du Core Stability in situ sur des athlètes de haut niveau.

La première phase visera à la détermination de l'influence du Core Stability sur i) le transfert de force et ii) sur la performance en aviron. 20 rameurs de haut niveau (niveau national et international) seront recrutés au travers de la structure du Pôle France Aviron à Nancy. Les athlètes réaliseront un test en situation écologique, à savoir une performance en aviron sur une distance de 500m, départ arrêté. Lors de ce test, le Core Stability, le transfert de forces et la performance de course seront mesurés. Le Core Stability sera évalué à la fois dans ses composantes biomécaniques cinématiques et neuromusculaires. L'analyse cinématique sera effectuée à l'aide de deux capteurs inertiels Captiv T-Sens Motion (TEA, Vandoeuvre-lès-Nancy, France), placés dans le dos entre les deux scapulas et sur l'extérieur de l'épine iliaque antéro-supérieure. Les paramètres cinématiques d'inclinaison du thorax et du pelvis dans les trois plans pourront alors être relevés. L'analyse électromyographique sera effectuée à l'aide de 6 capteurs d'électromyographie de surface (EMG) Trigno™ Wireless EMG System (Delsys, Natick, MA, USA) placés au niveau des muscles rectus abdominis, obliquus externus abdominis et erector spinae pour le tronc (à droite et à gauche à chaque fois). L'analyse permettra d'obtenir une enveloppe du signal et la latence d'activation des différents muscles par rapport aux phases de propulsion et de retour, ce qui permettra d'identifier les stratégies utilisées. Le transfert de force de la barre de pieds vers la dame de nage sera mesuré grâce à

des capteurs de force spécifiques pour l'aviron (Peach Innovations Ltd., Cambridge, Angleterre). Cette mesure permettra l'évaluation des forces verticales et horizontales au niveau de la barre de pieds et de manière synchrone la force efficace produite au niveau de la dame de nage. Le rapport entre ces deux composantes de la force permettra de déterminer le transfert de force. Les paramètres de la performance en aviron, que sont la vitesse de bateau et le temps de course seront mesurés par le GPS intégré au bateau.

Lors du test de 500m, les variables du Core Stability, le transfert de force ainsi que les variables de performance seront évalués pour chaque coup d'aviron puis moyennés sur 10 cycles à différentes distances de course (100m, 200m, 300m, 400m et 500m). L'analyse des régressions linéaires entre les paramètres du Core Stability (cinématiques et neuromusculaires) et ceux du transfert de la force aux différentes distances de l'épreuve de 500m permettra de vérifier l'hypothèse principale d'une relation existante entre le Core Stability et le transfert de force en aviron. L'analyse des régressions linéaires entre les paramètres du Core Stability et ceux de la performance aux différentes distances de l'épreuve de 500m permettra de vérifier l'hypothèse principale d'une relation existante entre le Core Stability et la performance en aviron.

La seconde phase visera à déterminer l'aménagement optimal de l'embarcation en fonction des caractéristiques morphologiques et techniques des rameurs. Cette étape de développement industrielle nécessitera la mise en adéquation des principaux points de liaison de l'athlète avec le bateau (barre de pieds, siège, dames de nage) avec les données anthropométriques des athlètes et leurs comportements techniques en lien avec les résultats du Core Stability. Pour cette étude, les caractéristiques anthropométriques des athlètes seront mesurées à l'aide du marbre ML Size (MorphoLogics). Cet outil de haute précision permet la lecture de l'ossature du corps humain à l'aide de capteurs mobiles, connectés par transmission bluetooth. Différentes mesures anthropométriques : stature, largeur d'épaule, hauteur d'épaule, longueur des bras, largeur du bassin, longueur des fémurs seront réalisées, auxquelles s'ajoutent deux mesures du pied. Le système électronique intégré dans le ML Size, permet la numérisation des mesures. La confrontation des données anthropométriques avec les données du Core Stability et du transfert de force de la première phase permettra un développement individualisé de l'aménagement du bateau en vue d'optimiser la performance.

A plus long terme, ce projet pourrait ouvrir la voie à des coopérations avec la Fédération Française d'Aviron, et l'Institut National du Sport, de l'Expertise et de la Performance (INSEP) afin de transférer ces nouvelles connaissances sur le Core Stability à la

compréhension des possibilités d'ergonomie du bateau, ainsi qu'une meilleure compréhension du risque de blessure au niveau de la charnière lombo-sacrée en aviron.

IV.3) A long terme : Analyse biomécanique des gestes répétés au travail et prévention des Troubles Musculo Squelettiques (TMS)

Ce projet s'inscrit tout d'abord dans le cadre de l'évolution du Master STAPS Mention Ingénierie et Ergonomie de l'Activité Physique, parcours-type « Métrologie, Performance, Santé », dans lequel je porte une responsabilité pédagogique (2018-2023). De plus, ce projet sera cohérent par rapport à la création à venir de l'Unité de Prestation et de Services « ErgoSim », dont le but sera de proposer des solutions de Métrologie, de Modélisation et de Simulation dans une approche ergonomique au service des entreprises et autres collectivités. Ces solutions ergonomiques se voudront préventive et auront, à ce titre, comme objectif une meilleure adaptation du travail à l'homme. J'exercerai les fonctions de directeur scientifique dans cette UP&S ErgoSim, qui sera rattachée à l'UFR STAPS – Faculté des Sciences du Sport de Nancy en lien avec l'équipe d'accueil EA 3450 DevAH « Développement – Adaptation – Handicap ».

Les TMS recouvrent un large éventail de pathologies touchant les tissus mous à la périphérie des articulations et de la colonne vertébrale, dont les plus représentées sont les lombalgies, les tendinites et les syndromes du canal carpien. En France, les TMS sont la première cause de maladie professionnelle. Ils représentent 80% des maladies professionnelles pour les salariés du régime général. Selon les années, entre 35 000 et 40 000 nouveaux cas de TMS ont été indemnisés, qui s'ajoutent à ceux des années antérieures, et le coût moyen direct d'un TMS pour l'entreprise, avec ou sans arrêt de travail, est estimé à 21 300 €. Au total, l'ensemble des TMS indemnisés a engendré la perte de 8 à 9 millions de journées de travail et environ 800 millions d'euros de frais couverts par les cotisations des entreprises. Même si la santé des salariés et les caractéristiques propres à chaque individu (âge, sexe, corpulence...) peuvent moduler le niveau d'exposition, les récentes études montrent que les TMS résultent d'une combinaison de plusieurs facteurs de risque, que l'on regroupe généralement dans trois grandes catégories : 1) les facteurs biomécaniques (répétitivité des gestes, postures et angles articulaires extrêmes...), 2) les facteurs organisationnels (organisation du travail, environnement des postes de travail) et 3) les facteurs psychosociaux (latitude décisionnelle / contrainte organisationnelle, modalité relationnelle et communicationnelle, précarité...). Dans

chaque entreprise, la combinaison des facteurs de risque peut être différente, ce qui exige une appréhension individualisée et personnalisée des situations de travail.

L'objectif de ce projet est de développer différents protocoles d'évaluation des contraintes biomécaniques en situation écologique à partir des outils classiques de mesure cinématique, cinétique et électromyographique déjà présentés dans ce travail d'HDR. De plus, l'analyse des contraintes musculo-squelettiques utilisera des modèles du même nom sous OpenSim (Blache, Begon, Michaud, Desmoulins, Allard, & Dal Maso, 2017). A terme, une meilleure adaptation du travail à l'homme aura pour résultat le bien-être des opérateurs, leur confort, leur santé mais aussi l'efficacité de leurs conduites opératoires et cognitives. Elle se traduit par conséquent par un gain de productivité de l'entreprise et une amélioration des conditions de travail des salariés en direction de la préservation de leur bien-être et de leur santé.

V. BIBLIOGRAPHIE

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VI. FINANCEMENTS

Type de financement	Projet	Montant	Universités	Années
Appel à projets	Enveloppe « Trajectoire » Pôle BMS	5.460€	Univ. Lorraine	2018
Contrat doctoral	Thèse Y. Duchène		Univ. Lorraine	2017-2020
	Adidas_Apparel_49	9.800€	Univ. Freiburg	2013
	Adidas_Apparel_48	6.600€	Univ. Freiburg	2012
	Adidas_Apparel_46	8.700€	Univ. Freiburg	2012
	Adidas_Football_45	4.200€	Univ. Freiburg	2012
	Adidas_Football_44	6.800€	Univ. Freiburg	2012
	Adidas_Football_43	26.500€	Univ. Freiburg	2011
Prestation de service	Adidas_Football_41	23.500€	Univ. Freiburg	2011
	Adidas_Core Skill_42	4.500€	Univ. Freiburg	2011
	Adidas_Football_38	13.000€	Univ. Freiburg	2011
	Adidas_Football_37	5.700€	Univ. Freiburg	2010
	Adidas_Core Skill_39	4.000€	Univ. Freiburg	2010
	Adidas_Running_35	37.200€	Univ. Freiburg	2010
	Adidas_Core Skill_36	11.100€	Univ. Freiburg	2010

VII. LISTE DE PUBLICATIONS

VII.1) Publications

Publications internationales :

23 articles acceptés dans une revue indexée **SCI** (9 en tant que premier auteur).

* ces articles sont les plus significatifs de ce travail d'HDR et une copie est disponible ci-après.

23 Mornieux, G., Hirschmüller, A., Gollhofer, A., Südkamp, N. P., & Maier, D. (2018). Multimodal assessment of sensorimotor function in patients with untreated anterior shoulder instability and asymptomatic handball players. *The Journal of Sports Medicine and Physical Fitness*, 58, 472-479.

22* Mornieux, G., Weltin, E., Pauls, M., Rott, F., & Gollhofer, A. (2017). Influence of a full body compression suit on trunk positioning and knee joint mechanics during lateral movements. *Journal of Applied Biomechanics*, 33, 261-267.

21* Weltin, E., Gollhofer, A., & **Mornieux, G.** (2017). Effects of perturbation or plyometric training on core control and knee joint loading in women during lateral movements. *Scandinavian Journal of Medicine & Science in Sports*, 27, 299–308.

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Influence of a Full-Body Compression Suit on Trunk Positioning and Knee Joint Mechanics During Lateral Movements

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Trunk positioning has been shown to be associated with knee joint loading during athletic tasks, especially changes of direction. The purpose of the present study was to test whether a full-body compression suit (FBCS) would improve trunk positioning and knee joint control during lateral movements. Twelve female athletes performed lateral reactive jumps (LRJ) and unanticipated cuttings with and without the customized FBCS, while 3D kinematics and kinetics were measured. FBCS did not influence trunk positioning during LRJ and led to increased trunk lateral lean during cuttings ($P < .001$). However, while wearing FBCS, knee joint abduction and internal rotation angles were reduced during LRJ ($P < .001$ and $P = .013$, respectively), whereas knee joint moments were comparable during cuttings. FBCS cannot support the trunk segment during unanticipated dynamic movements. But, increased trunk lateral lean during cutting maneuvers was not high enough to elicit increased knee joint moments. On the contrary, knee joint abduction and internal rotation were reduced during LRJ, speaking for a better knee joint alignment with FBCS. Athletes seeking to improve trunk positioning may not benefit from a FBCS.

Keywords: cuttings, perturbation, trunk lateral lean, knee abduction moment, agility

Lateral movements with changes of direction are common in many team sports such as soccer or handball. During cutting maneuvers, the athlete performs a complex dynamic task by quickly changing their direction of movement while securing their balance. Knee joint control is an essential requirement during such movements to prevent misalignments in the lower extremities with the consequence that the knee may be injured.^{1,2} Noncontact anterior cruciate ligament (ACL) injury mechanism during cutting maneuvers typically involves knee external flexion moment combined with dynamic valgus rotation with the body weight shifted over the injured leg.² Trunk positioning during lateral movements is of interest as it would influence the location of the body weight, and potential knee joint loads.^{3,4} Hewett and Myer³ proposed a knee injury model that accounted for both the trunk and hip joints, because increased trunk lateral lean is reportedly associated with larger external knee abduction moments. Greater lateral trunk position has been shown during ACL injury situations in female athletes.⁵ Different laboratory studies using cutting maneuvers and lateral reactive jumps paradigms have demonstrated that increased trunk lateral lean is positively associated with increased knee abduction moments.⁶⁻⁸ Further, increased knee joint loading has been associated with increased trunk rotation away from the stance limb.^{8,9} For instance, higher knee internal rotation moment was found for trunk rotation in the opposite direction to the cutting maneuver.⁸ Accordingly, any improvement of trunk positioning is of interest with regard to knee joint loading. This would be especially true for female athletes, whose incidence of ACL

injuries in sports such as soccer and basketball is much higher than male athletes.^{10,11}

In sport sciences, the influence of compression garments, providing passive support to different body segments during exercise, has been extensively studied.¹² It has been showed that compressive and elastic garments could reduce the range of motion of the joint.^{13,14} Compression shorts reduced the active range of motion during hip flexion in a nonathletic task¹³ and hip range of motion during a 60-m sprint.¹⁴ Wearing shorts with directional compression was associated with reduced hip adductor activity during cutting manoeuvres.¹⁵ In this latter study, the use of diagonal bands of fabric that only stretch along their length increased the compression level along the line of action of hip adductors. Therefore, apparel, through its compression and elasticity properties, could be used to passively support body segments and joints during various tasks. A compression suit that would functionally support the trunk over the pelvis and leg could be a solution to avoid trunk positioning in the opposite direction to the intended movement. Such apparel would work as external trunk stabilizers to improve trunk positioning during various tasks, especially lateral movements, and subsequently possibly reduce knee joint loading.

In sports, any improvement of movement control would further be relevant if associated with enhanced performance. However, when considering whole-body dynamic tasks, wearing a compression suit did not influence performance in cricket players¹⁶ or endurance performance during running.¹⁷ Only 1 study has shown possible benefits of a whole-body compression garment on performance during high-intensity intermittent exercises, mainly because of better muscle oxygenation.¹⁸ Therefore, athletic performance is not likely to be increased when using garments.

The aim of this present study was to test whether compression garments can improve trunk and hip positioning and resultant knee joint loading, while not decreasing performance during dynamic lateral sporting movements. For that purpose, a full-body compression suit with functional elastic straps was designed to improve trunk positioning and knee joint control during lateral movements.

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We tested the hypotheses that (i) trunk and hip joint angles would be reduced with the compression suit, (ii) knee joint abduction moment would be reduced with the compression suit, and (iii) exercise performance level would be maintained while wearing the compression suit.

Methods

According to other studies evaluating comparable dependent measures during unanticipated lateral movements¹⁹ or the influence of compression shorts on sports performance,¹³ a sample size of 13 participants was estimated relevant for statistical analysis. The rationale for focusing on females was their relative higher risk of sustaining noncontact ACL injuries.¹⁰ Among the 13 female athletes recruited to participate in the study, 1 was injured during a training session with her team during the investigation period and could not take part in the measurements. Therefore, 12 female athletes (body height = 171 ± 4 cm; body mass = 59 ± 4 kg; age = 22 ± 2 years) who were familiar with lateral movements (soccer, basketball, and handball) completed the entire study. Participants did not have a previous history of serious knee injury or any current knee pain. Written consent was obtained prior to study participation. The study was approved by the local ethics committee (approval 72/14) and was conformed to requirements stipulated in the Declaration of Helsinki.

During a first visit, participants' body dimensions were scanned (BodyKinectizer, adidas IT Innovation, Herzogenaurach, Germany), in order to customize the full-body compression suit. During a second visit at our laboratory, participants performed different lateral movements and performance tests under 2 different apparel conditions, either while wearing the customized full-body compression suit (FBCS) with 2x thermoplastic polyurethane (TPU) layers or with standard sport shorts and bras (Control). FBCS was a prototype specially designed for the present study in order to provide support for the trunk over the pelvis segment in 3 dimensions. For that purpose, a mean compression level of 5.2 mmHg was achieved and the 2x TPU layers design linked the upper body with lower limbs across the pelvis in the back (Figure 1). The compression level was first determined for elastic and TPU fabrics separately, according to their relative stretch and then averaged over the whole suit. At specific points, the compression level was even greater, for instance at the low waist and thigh lines (8.3 and 6.7 mmHg, respectively). By determining the compression exactly and creating a customized suit for each subject, identical apparel conditions were guaranteed for all subjects, and the possible effectiveness of the compressive suit was measurable.²⁰ To determine to what extent wearing FBCS would influence trunk range of motion, trunk passive flexibility in frontal and sagittal planes was measured according to the literature.²¹ In addition, trunk rotation passive flexibility was evaluated while participants lay on their backs with the right bent knee crossing over the thigh of the left extended leg. Participants remained relaxed and kept their back and both shoulders in contact with the floor, while the distance between the floor and knee was measured. Reduced mean trunk passive flexibility over all participants was obtained with FBCS compared to Control for flexion (2.2 cm) and rotation (2 cm), but was comparable for lateral flexion.

The 2 conditions were randomly assigned over all participants. According to pilot measurements, participants were given 30 minutes to become accustomed to the suit. During this period, standardized exercises were carried out based on runs, jumps, and cuttings. When subjects started the measurement with FBCS,



Figure 1 — Participant wearing the customized full-body compression suit (FBCS) with 2x thermoplastic polyurethane layers (TPU). The FBCS design provides support for the trunk over the pelvis segment through the 2x TPU layers that link the upper body with lower limbs across the pelvis in the back (right). The customized apparel production ensures a perfect fitting for each participant.

another 30 minutes practice was given to them prior to the Control condition.

Firstly, participants performed perturbed lateral reactive jumps (LRJ) to focus on the influence of FBCS on movement control during a functional task associated with high trunk lateral lean in the frontal plane. From a standing position, subjects performed a lateral reactive movement by jumping to their left side and then pushing off with their left leg to get back to their original position. Jump distance was set at 124% of the individual leg length. In order to induce lateral perturbations, subjects performed LRJ on a motor-driven plate, which could translate in the horizontal (medio-lateral) direction. The specially designed motor-driven plate (University of Sport Science, Freiburg, Germany) comprised 2 independently moving plates ($1.15 \text{ m} \times 0.3 \text{ m}$), each driven by a separate motor (AKM51G, Danaher Motion, Düsseldorf, Germany). Details of this material have been previously described in the literature.²² LRJ were performed, either without any perturbation (stable condition), or with a perturbation to the left (sliding condition) or to the right (counteracting condition) as soon as the foot made initial contact with the plate. Each perturbation consisted of a 10-cm translation in either direction over 250 milliseconds, inducing a peak velocity of $0.55 \text{ m} \cdot \text{s}^{-1}$ after 120 milliseconds. Unpredictability was introduced using these 3 different landing modalities (stable, sliding, counteracting) that were unknown prior to movement execution. Perturbation was triggered by breaking a light beam (M18 series, Panasonic Electric Works Europe AG, Holzkirchen, Germany), 0.04 m above the plate, to compensate for the delay in activation of the motor-driven plate.

Secondly, participants were asked to perform either cutting maneuvers to 45° , crossovers to -20° , or straight runs on a force plate (BP600900, AMTI®, Watertown, USA), after approaching the force plate (FP) at $4 \pm 0.2 \text{ m} \cdot \text{s}^{-1}$. The direction of the movement was indicated in a randomized order by a light signal occurring

650 milliseconds before the left foot contacted the FP. The visual light stimulus was triggered automatically during the approach run via a light switch (M18 series, Panasonic Electric Works Europe AG, Holzkirchen, Germany) placed 3.7 m from FP. According to a light cue delay set individually, the 650-millisecond time could be achieved. The 3 lights were placed at eye level in the directions of the 3 tasks, 7 m away from the FP. The 45° cutting maneuver to the right was the only movement to be further analyzed due to its relevance when investigating performance and injury aspects of changing direction tasks.

Finally, 3 different performance tests were conducted to evaluate agility with the 5-10-5 yards test, jump height, ie, leg muscle power, with countermovement jumps (CMJ) and core strength with a cable lift test. This latter test targets the action of hips, torso rotators, upper back, chest, and shoulders while turning the trunk away from the strength machine as the hands first pull the rope cable handle toward the chest and then push up and away. The rationale for these tests is the importance of agility, leg muscle power, and core strength for high performance during soccer, basketball, or handball games. The good reliability of jump height measurements during CMJ has been reported in the literature (intra-class correlation coefficient [ICC] = 0.9),²³ as well as for the 5-10-5 yards agility test (ICC > 0.87).²⁴ The time to accomplish the agility test was measured manually. Two CMJ were performed on the FP. The cable lift test was conducted twice at 70% of 1-repetition maximum.

All participants were familiar with such movements due to their sport practice and were given enough practice trials to feel comfortable with all task modalities prior to the measurements.

Three-dimensional leg kinematics were recorded using reflective skin markers (Ø 14 mm) placed on anatomical landmarks of each subject's trunk (3 markers), pelvis (4 markers), left thigh (3 markers), shank (4 markers), and foot (6 markers). Markers were captured with a 12-camera motion analysis system (Vicon MX™, VICON Motion Systems Ltd., Oxford, UK) with a sampling frequency of 200 Hz. A static trial with the subject standing in a predefined neutral position was used to define the kinematic model and to calculate segment lengths and joint centers. The kinematic modeling was performed with the motion analysis software (Vicon Nexus 1.6.1, VICON Motion Systems Ltd., Oxford, UK). Joint kinematics in 3 rotational degrees of freedom were determined using an yxz Euler rotation sequence of the respective segment coordinate system according to the global reference system defined in Vicon. This rotation sequence was around the flexion/extension axis first, then around the abduction/adduction axis, followed by the rotation around the external/internal axis. This marker placement and kinematic modeling were used in previously published articles.^{7,19}

Marker trajectory and ground reaction force signals were both filtered with a low-pass Butterworth filter (fourth-order, 15-Hz cut-off frequency) prior to calculating external joint moments with a standard inverse dynamics approach. Three-dimensional kinematic data of the trunk segment and hip joint were analyzed to understand the influence of FBSC on trunk positioning. Three-dimensional kinematics and kinetics of the knee joint were further analyzed to determine the influence of FBSC on knee joint stability and loading. Both kinematics and kinetics were measured during cutting maneuvers, but only kinematics was observed during LRJ. These variables were analyzed at the time of their peak values during the first half of ground contact during LRJ, to give some insights into how the apparel condition influences the range of motion of the different body segments. However, during cutting maneuvers, the different variables were determined at the time of

peak knee abduction moment, as this is a relevant parameter associated with knee injury.²⁵

Trunk lateral lean and trunk axial rotation were defined relative to the global coordinate system, ie, deviations from the vertical axis and from the anterior-posterior axis, respectively. Both were positive for an orientation in the new movement direction (ie, to the right direction). Hip rotation, knee rotation, and knee rotation joint moment were defined as positive for an external rotation.

The time to accomplish the 5-10-5 yards test served as outcome measure for agility. CMJ height was calculated based on the flight time method. During cable lift testing, the vertical acceleration signal of the accelerometer (University of Sport Science, Freiburg, Germany) placed on the top of the weight stack was filtered with a low-pass Butterworth filter (fourth-order, 20-Hz cut-off frequency) and then integrated to obtain velocity. Based on Newton's second law, lift force was calculated. Finally, power was determined by means of the product of force and velocity.

The selected parameters were averaged across 8 trials for the different LRJ modalities and the 45° cutting maneuver, while the time during the 5-10-5 yards test was reported, and the highest CMJ and mean peak power value over 2 cable lift trials were calculated. These parameters served as the basis for the statistical analysis (Statistica 12, StatSoft, Inc., Tulsa, OK, US). All results are presented as group mean ± standard deviation (SD). After having confirmed that data followed a normal distribution (Shapiro-Wilk W), sphericity was verified by means of the Mauchly's test. The influence of the apparel condition on dependent variables for LRJ modalities was calculated using a 2-way analysis of variance (general linear model) with repeated measures to test for main effects of and interactions between apparel (Control vs FBSC) and LRJ modality (stable vs sliding vs counteracting). If a statistically significant LRJ modality main effect was observed, Bonferroni-corrected paired comparisons determined where the differences occurred. The influence of the apparel condition (Control vs FBSC) on the dependent variables for the cutting maneuver and performance tasks was analyzed using paired 2-tailed *t*-tests. The magnitude of the changes was assessed with effect size (ES) and 95% confidence interval (CI). The ES was evaluated by means of partial eta squared (η_p^2) values and were considered small (.01–.06), medium (.06–.14), and large (> .14), according to Cohen.²⁶ The level of significance was set at $P < .05$.

Results

There was a significant main effect of the apparel condition on hip rotation, knee abduction, and rotation during LRJ (Table 1). Hip joint internal rotation was increased with FBSC ($P = .019$), while knee joint abduction and internal rotation were reduced ($P = .002$ and $P < .001$, respectively). Statistical analysis revealed a significant main effect of the LRJ modality on trunk forward lean and trunk lateral lean ($P = .025$ and $P = .003$, respectively), hip flexion, abduction and rotation ($P < .001$ for all), and knee flexion and rotation (both $P < .001$). However, no interaction effect (apparel condition × LRJ modality) was reported. Specifically, pairwise comparisons revealed that stable, sliding, and counteracting modalities yielded significantly different values from each other for trunk lateral lean ($P < .004$ for all), hip flexion ($P < .001$ for all), hip rotation ($P < .003$ for all), knee flexion ($P < .001$ for all), and knee rotation ($P < .003$ for all). Trunk flexion was significantly greater during counteracting than both stable ($P < .02$) and sliding ($P < .001$). Hip abduction was significantly greater during sliding than both stable ($P < .001$) and counteracting ($P < .001$).

Table 1 Mean \pm SD values of the different kinematic variables during lateral reactive jumps (LRJ) performed in control and full-body compression suit (FBCS) conditions

Variable	Control (Pooled LRJ)	FBCS (Pooled LRJ)	P Value	95% CI	ES (η_p^2)
Trunk forward lean ($^\circ$)	32.0 \pm 6.2	31.6 \pm 6.8	.66	[-1.7, 2.5]	.02
Trunk lateral lean ($^\circ$)	-7.5 \pm 5.4	-7.9 \pm 6.1	.63	[-1.4, 2.2]	.02
Trunk right axial rotation ($^\circ$)	4.6 \pm 9.4	6.7 \pm 9.3	.10	[-4.7, 0.5]	.23
Hip flexion ($^\circ$)	50.6 \pm 8.9	51.1 \pm 13.3	.84	[-5.5, 4.5]	< .01
Hip abduction ($^\circ$)	35.5 \pm 4.7	35.7 \pm 5.7	.84	[-2.5, 2.0]	< .01
Hip external rotation ($^\circ$)	-5.0 \pm 6.5	-11.1 \pm 6.2*	.019	[1.2, 10.9]	.41
Knee flexion ($^\circ$)	59.0 \pm 5.3	59.9 \pm 5.1	.10	[-2.0, 0.2]	.23
Knee abduction ($^\circ$)	10.3 \pm 6.7	5.3 \pm 3.9**	.002	[2.3, 7.5]	.61
Knee external rotation ($^\circ$)	-13.5 \pm 7.4	-8.6 \pm 6.5***	< .001	[-7.3, -2.6]	.66

Note. The influence of the apparel condition is reported using *P*, 95% confidence interval (95%CI) and effect size (ES) with partial eta squared (η_p^2) values. Significant differences between Control and FBCS with **P* < .05, ***P* < .01, and ****P* < .001, respectively.

Table 2 Mean \pm SD values of the different kinematic and joint moment variables during cutting maneuvers performed in control and full-body compression suit (FBCS) conditions

Variable	Control	FBCS	P value	95% CI	ES (η_p^2)
Trunk forward lean ($^\circ$)	12.6 \pm 7.8	13.9 \pm 5.6	.22	[-3.6, 0.9]	.14
Trunk lateral lean ($^\circ$)	-5.1 \pm 3.7	-8.7 \pm 5.0***	< .001	[2.2, 5.0]	.74
Trunk right axial rotation ($^\circ$)	-8.5 \pm 8.4	-7.8 \pm 7.5	.61	[-3.4, 2.1]	.02
Hip flexion ($^\circ$)	51.6 \pm 6.7	49.3 \pm 10.6	.35	[-2.9, 7.5]	.08
Hip abduction ($^\circ$)	8.4 \pm 8.9	7.7 \pm 5.9	.59	[-2.2, 3.7]	.03
Hip external rotation ($^\circ$)	-4.2 \pm 7.3	-9.8 \pm 4.4**	.003	[2.4, 8.8]	.57
Knee flexion ($^\circ$)	28.3 \pm 5.0	28.0 \pm 6.1	.56	[-0.9, 1.6]	.03
Knee abduction ($^\circ$)	4.7 \pm 3.1	2.7 \pm 3.1	.054	[-0.1, 4.1]	.30
Knee external rotation ($^\circ$)	-7.4 \pm 6.6	-3.4 \pm 5.2	.052	[-8.1, 0.1]	.30
Knee flexion moment (Nm \cdot kg $^{-1}$)	0.16 \pm 0.65	0.20 \pm 0.75	.74	[-0.33, 0.24]	.01
Knee abduction moment (Nm \cdot kg $^{-1}$)	1.31 \pm 0.46	1.20 \pm 0.46	.30	[-0.12, 0.35]	.1
Knee extension rotation moment (Nm \cdot kg $^{-1}$)	-0.13 \pm 0.11	-0.16 \pm 0.12	.42	[-0.06, 0.13]	.06

Note. The influence of the apparel condition is reported using *P*, 95% confidence interval (95%CI) and effect size (ES) with partial eta squared (η_p^2) values. Significant differences between Control and FBCS with **P* < .01 and ****P* < .001, respectively.

During the cutting maneuver, trunk forward lean and axial rotation were not influenced by the apparel condition at the time of peak knee abduction moment, while trunk lateral lean was significantly increased by FBCS (*P* < .001). Hip joint internal rotation was higher for FBCS than Control (*P* = .003). Knee joint angles and moments, especially the knee abduction moment, were not influenced by the apparel condition (Table 2). The ensemble averages of trunk lateral lean and knee joint abduction moment over the stance phase of the cutting maneuver are presented (Figure 2).

Performance parameters were not significantly influenced by the apparel condition. The sprint time measured during the 5-10-5 yards test was comparable between the 2 apparel conditions (5.6 \pm 0.3 s for both conditions; *P* = .90, 95% CI [-0.15, 0.13], η_p^2 = .002). CMJ height was not significantly different between Control and FBCS conditions (0.28 \pm 0.05 m for both; *P* = .40, 95% CI [-0.01, 0.01], η_p^2 = .07). Mean peak power produced by the participants during the cable lift did not differ significantly (*P* = .55, 95% CI [-6, 11], η_p^2 = .03) when performing the movement in the Control condition (282 \pm 69 W) or with FBCS (280 \pm 68 W).

Discussion

The main findings of the study were as follows: (1) FBCS did not reduce trunk angles during lateral movements, with even an increased trunk lateral lean during cutting maneuvers; (2) hip internal rotation was increased with FBCS; (3) knee joint moments were not reduced with FBCS; and (4) performance level was kept at the same level.

Perturbed lateral reactive jumps yielded significantly different segment and joint configurations at the time of their peak values. A detailed analysis of the influence of LRJ modality at the time of initial contact has already been provided in the literature.²⁷ During perturbed lateral reactive jumps, trunk positioning was not influenced by the compression garment. It seems that despite a suit design enabling a reduced passive trunk range of motion, FBCS cannot support the trunk segment during an active dynamic movement. Thus, the trunk segment was not supported either during cutting maneuvers. On the contrary, trunk lateral lean was even significantly increased by 4 $^\circ$ while wearing FBCS during 45 $^\circ$ cutting maneuvers. While all participants showed greater trunk lateral lean with FBCS than without apparel, 5 subjects reached

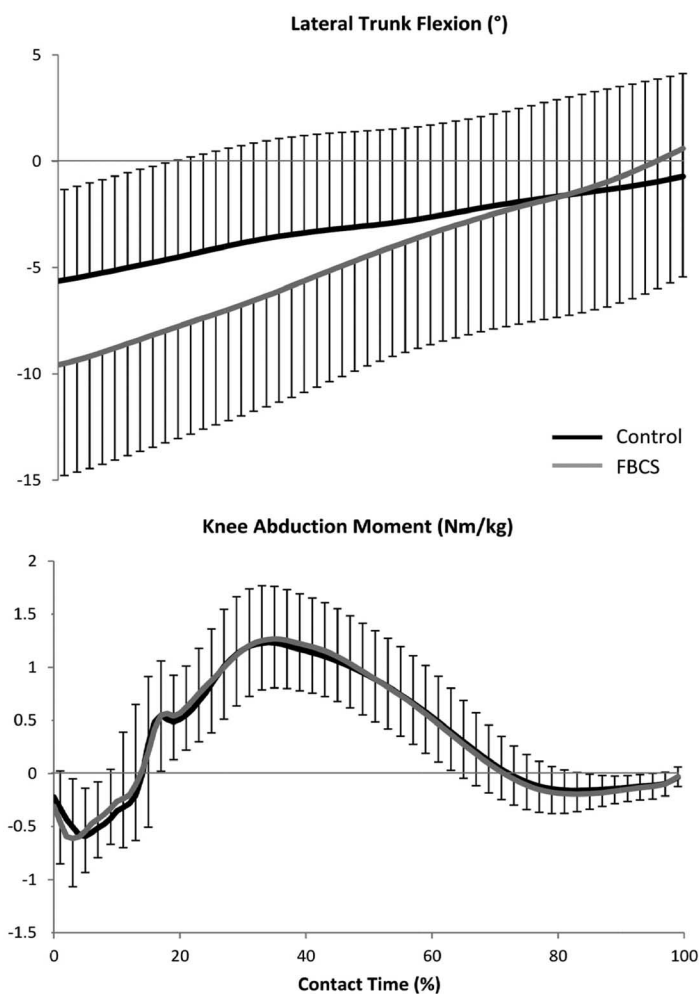


Figure 2 — Ensemble averages of lateral trunk flexion in the frontal plane and knee joint abduction moment for Control (black line) and full-body compression suit (FBCS, gray line) during the foot contact phase of the cutting maneuver.

particularly large values around 15° with FBCS. Although literature demonstrated that compression garments could reduce the active range of motion of the hip joint during a nonathletic task¹³ and during sprinting,¹⁴ a more neutral trunk positioning through apparel could not be achieved during unanticipated lateral movements. It might be possible that participants would rely on the support of the suit according to the compression and elasticity properties felt by the participants, thereby reducing their active trunk muscle control. But the lack of apparel support would bring participants to this higher trunk lateral lean in the frontal plane during cutting maneuvers.

The specific FBCS linked shoulders with knees across the pelvis in the back thanks to the elastic straps design. But, at the hip joint level, only hip internal rotation was altered. The hip internal rotation observed in the present study is consistent with literature during lateral reactive jumps²⁷ and cutting maneuvers.²⁸ However, increased peak internal rotation values with FBCS might rather reflect a lack of strength in the external hip rotators in female athletes,²⁹ exacerbated by the additional tension of the compression garment.

It appears that external passive support with the aim of enhancing trunk positioning during dynamic lateral movements

classically encountered during team sports is difficult to achieve with apparel. At this point, further improvements to the design of the compression suit does not seem possible as several years of research and development were needed to yield this current comfortable full-body suit, which enables reduced passive trunk flexibility through the combined compression and elasticity properties. In addition, garments were customized to each participant in the present study. Further increasing the compression level of the suit to further support body segments, ie, by reaching typical values of compression socks > 15 mmHg,³⁰ does not seem possible given that (i) the straps used already increase the stiffness of the suit and (ii) the higher compression level would reduce the comfort level necessary to perform whole-body dynamic movements.

Trunk and hip control alterations with FBCS could have consequences at the knee joint level. Indeed, higher trunk lateral lean during cutting maneuvers while wearing FBCS might be associated with increased knee joint loading.^{6,7} In the present study, higher trunk lateral lean had no deleterious influence on the knee joint, as no significant difference in knee abduction moment was evident between FBCS and control conditions. Increased peak internal hip rotation angles with FBCS might reflect a lack of strength in the external hip rotators²⁹ and further a critical knee joint position (position of no return), where the hip moves more rapidly into adduction and internal rotation and the knee collapses in valgus.³¹ But the consequences at the knee joint level during lateral reactive jumps were rather a reduction of the abduction and internal rotation angles whatever the landing condition with FBCS, which speaks in favor of better knee joint alignment. Thus, alteration of trunk positioning during lateral movements with FBCS was not associated with changes in knee joint stability.

This result does not support the theoretical model proposed by Hewett and Myer³ and is not in agreement with experimental results, where increased trunk lateral lean was associated with increased knee joint abduction moment.^{6,7} This finding underlines the fact that trunk positioning only partially influences knee joint loading, and that relatively small trunk angle variations (4°), albeit significant, may not always have an effect on knee abduction moment, especially when considering the large variability of this knee joint loading parameter.

It is worth noticing that FBCS had no influence on any of the performance parameters. The same level of agility while wearing the compression suit or in the control condition during the 5-10-5 yards test is consistent with the literature.^{13,16,32} Although improvement of jumping performance with a compression garment has been reported in the literature,^{14,33} it is interesting to note that jump height was not impaired with the compression suit targeting trunk and pelvis positioning rather than the lower limbs in the present study. Finally, core strength during the cable lift task was kept at the same level when performed with FBCS. Accordingly, FBCS can be used by athletes without risk of compromising performance.

This study has some limitations that need to be considered. The moderate reliability of knee kinetics during dynamic tasks such as cutting tasks would suggest the need for a greater sample size than 12 participants.³⁴ In the present study, measured variables were assessed over 8 trials to ensure representative values for each participant. But, the intersubject variability was pretty high for joint kinetics and the associated statistical power low (< 0.2). Thus, future studies would possibly need a larger sample size to tease out effects on knee joint kinetics or at least enable the ability to discuss the lack of difference in knee joint loading. Another potential limitation was the use of a cable lift task. While this test seems to be

relevant to evaluate core strength, the validity and reliability of this measure would need to be assessed to determine its measurement error. This will allow evaluating the importance of observed differences from a statistical perspective as well as in terms of its practical implication.

Even if compression and elasticity properties were combined with this customized full-body compression suit, trunk kinematics could not be favorably influenced. Achieving external support of trunk positioning in female athletes through apparel seems to be difficult during unanticipated lateral movements. However, a full body compression suit did not impair any of the performance parameters and even yielded a better alignment of the knee during lateral reactive jumps.

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Effects of perturbation or plyometric training on core control and knee joint loading in women during lateral movements

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Deficits in trunk control are argued to increase the risk of knee injuries. However, no existing training program effectively addresses trunk control during lateral movements, such as cutting maneuvers. The purpose of this study was to investigate whether a combination of perturbation and plyometric training (PPT) would reduce trunk excursions against the new movement direction and reduce knee joint moments during lateral movements. Twenty-four active women participated in a RCT, where trunk and pelvis kinematics and knee joint moments were measured during lateral reactive jumps (LRJ) and unanticipated cutting maneuvers before and after a 4-week PPT program and compared to a control group. During

LRJ, trunk rotation away from the new movement direction was reduced ($P < 0.001$), while pelvis rotation toward the new direction was increased ($P = 0.006$) after PPT. Moreover, decreased knee extension moments ($P = 0.028$) and knee internal rotation moments ($P < 0.001$) were reported after both trainings. Additionally, PPT reduced trunk rotation by 7.2° during unanticipated cuttings. A 4-week PPT improved core control by reducing trunk rotation and reduced knee joint moments during LRJ. During training, perturbations should be introduced to improve core control during dynamic athletic movements, possibly reducing the risk of ACL injuries.

Most anterior cruciate ligament (ACL) injuries occur in non-contact situations, usually direction changes such as cutting or pivot turning, which are common in multidirectional sports, such as soccer, basketball, or handball. Female athletes have a substantially higher risk of injuring the ACL than their male counterparts participating in the same sports (Arendt et al., 1999; Agel et al., 2005). The mechanism underlying this gender disparity in ACL injury risk is most likely multifactorial in nature (Hewett et al., 2006). Among the risk factors discussed that contribute to the disparity in ACL injury incidence between men and women, biomechanical and neuromuscular control of the knee and core seems to play an important role.

Indeed, several studies have shown that women display knee and hip kinematics and kinetics in the frontal and transverse planes that are associated with increased injury risk in lateral movements such as cuttings (Beaulieu et al., 2009; Pollard et al., 2011) and single-leg squats (Graci et al., 2012). Additionally, there is a growing body of evidence that insufficient trunk control possibly increases the risk of ACL injuries (Hewett & Myer, 2011; Hughes, 2014). Analysis of actual injury situations (Hewett et al.,

2009) as well as laboratory studies (Jamison et al., 2012a; Weltin et al., 2015) have demonstrated that increased lateral trunk lean is positively associated with an ACL injury or increased knee abduction moments. Furthermore, several studies suggest that higher trunk excursions in the transversal plane are associated with increased knee joint loading (Dempsey et al., 2007; Frank et al., 2013). Moreover, prospective studies (Zazulak et al., 2007a, b) have shown that deficits in trunk neuromuscular control and proprioception are strong predictors of ACL injuries in women but not men. In light of this evidence, training interventions aimed at reducing knee joint injury risk should not only focus on the lower limb but also on core control, defined as the ability to limit the magnitude of the trunk excursions (Zazulak et al., 2007a) combined with a functional positioning of the pelvis (Powers, 2010), during athletic movements.

Existing neuromuscular training programs have been reported to effectively change movement patterns, thereby improving biomechanical risk factors and reducing the incidence of ACL injuries in female athletes (Hewett et al., 2005a; Alentorn-Geli et al., 2009a; Stojanovic & Ostojic, 2012; Stevenson et al.,

2015). Most of these training programs are multi-modal and include components of strength, balance, plyometrics, core, and technique.

Training programs addressing the development of motor programs specific to lateral movements should enable athletes to control trunk and pelvic motion together with lower limb alignment during athletic maneuvers (Weltin et al., 2016). It has been shown that a warm-up program targeting, for instance, core stability would reduce the rate of ACL injury in adolescent female football players (Walden et al., 2012). Moreover, several studies have reported the positive effects of core stability and trunk-focused neuromuscular training on lower limb biomechanics during jumping and landing tasks (Chappell & Limpisvasti, 2008; Pfile et al., 2013; Dempsey et al., 2014) or hip muscle strength during functional tasks (Myer et al., 2008). When considering studies that investigated lateral movements like fast changes of direction where injuries are likely to occur (Olsen et al., 2004), a training modifying cutting technique has been reported to improve trunk positioning and reduce knee joint loading during anticipated side-step cutting tasks (Dempsey et al., 2009). However, after a training intervention focusing on trunk stabilization, neither trunk control nor knee joint loading during cutting maneuvers could be improved relative to resistance training (Jamison et al., 2012b). These latter results might be explained by the use of static stabilization exercises, which are possibly not suitable for changing trunk control during highly dynamic sports maneuvers. Overall there is no existing training intervention, except the cutting technique modification training performed by Dempsey et al. (2009) that effectively improves trunk control during lateral movements and it is likely that this originates from the training paradigms. Combining plyometric and perturbation training could be an appropriate design, better targeting core musculature and therefore possibly enhancing trunk control during high dynamic movements.

Perturbation training effectively altered knee kinematics and muscle activity during perturbed gait in patients after ACL rupture (Chmielewski et al., 2005) as well as lower limb muscle activation patterns in healthy athletic women during perturbed gait, potentially reducing the risk of an ACL strain injury (Hurd et al., 2006). Perturbation training in this case involved the maintenance of lower extremity balance during single-legged stance while the support surface was perturbed (Chmielewski et al., 2005; Hurd et al., 2006). But the influence of perturbation training on the biomechanics of more dynamic movements such as cuttings and lateral reactive jumps (LRJs) as well as the influence on more proximal structures such as the trunk or the pelvis in an athletic population has not been tested

yet. To address this point, a specific training paradigm was created in the present study by combining perturbation and plyometric training (PPT). This novel approach should be more effective in engaging and strengthening core muscles and thereby altering whole-body biomechanics during lateral movements.

Therefore, the purpose of the study was to determine the effects of PPT on (a) core control and (b) lower limb biomechanics, during lateral jumps and cutting paradigms. We hypothesized that the intervention group would demonstrate decreased trunk excursions in the frontal and transverse plane away from the new movement direction, together with decreased knee joint moments.

Methods

Subjects

An *a priori* power analysis was conducted to estimate the sample size based on recent published data (Mornieux et al., 2014b). It was estimated that at least 12 participants per group were needed to achieve 80% statistical power with an alpha level of 0.05. This was adjusted to 14 participants per group to account for possible dropouts. This sample size is in line with other studies (Jamison et al., 2012b; Pfile et al., 2013). Twenty-eight females were initially enrolled in the study (Fig. 1). All participants were interviewed individually to assess eligibility for participation. Subjects had to meet the following inclusion criteria: at least 5 years of active participation in team sports with high exposure to running, jumping, and cutting (soccer, handball, and basketball), no additional core stability or balance training, and no history of ACL injury or neurological disorders. All participants trained at least three times a week (up to five times) and played at different levels (1st to 5th Division). All subjects were informed about the procedures, purpose and possible risks associated with the experimental setup and gave their written consent prior to testing. The experiments were run in accordance with the Declaration of Helsinki and were approved by the local ethics committee. The participants were matched for height, weight, age, and skill. Afterwards, one participant of a matched pair was randomly assigned either to 4 weeks of plyometric and perturbation training (PPT) or 4 weeks of plyometric training, which served as control (CON), while the other participant was enrolled in the other training group. Four participants withdrew from the study after being assigned to a training program: two due to knee ligament injuries during participation in a match, one because of a bruised thigh resulting in more than two missed training sessions, and one due to muscular problems. Accordingly, 12 subjects per group were included in the analysis (PPT, 21 ± 3 years, 170 ± 7 cm, 62 ± 6 kg, 11 ± 4 years team sport experience vs CON, 22 ± 2 years, 170 ± 7 cm, 61 ± 7 kg, 10 ± 4 years team sport experience).

Intervention

All participants took part in regularly scheduled season strength training, practices, matches, and tournaments, while training sessions were attended three times per week additionally. Modified plyometric training designed by Chimera et al. (2004) served as CON training (Table 1), whereas different variations of LRJs were used for the PPT training (Table 2,

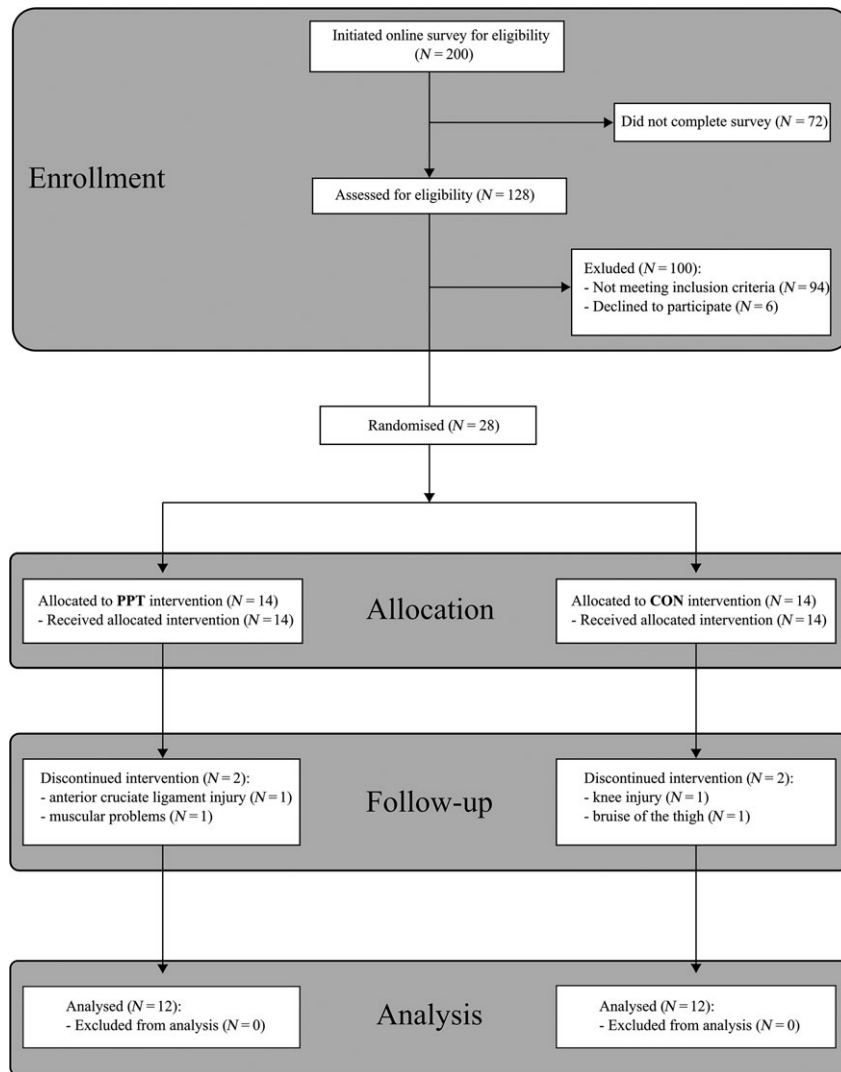


Fig. 1. Flow diagram of subject participation throughout all stages of the study.

see video appendices). LRJs were defined by jumping from an upright position over a predefined distance with one leg before pushing off with the other leg to get back to the starting position. Different variations of the LRJs were used during the training, i.e., either on the ground or performed on a custom-built motor-driven landing platform to induce perturbation upon landing. The custom-made motor-driven platforms (University of Freiburg, Germany) used in the study are made up of two independently moving plates (1.15 m × 0.30 m), each driven by a motor. A detailed description of the device has already been published (Mornieux et al., 2014b). While jumping on this platform, participants could experience different landing conditions at initial contact, i.e., (a) the landing platform moved away (sliding); (b) the landing platform moved toward the participant (counteracting); or (c) the landing platform did not move (stable). During the perturbation, the platform typically moved over 250 ms and reached a peak velocity of 0.55 m/s after 120 ms. Moreover, uncertainty could be introduced by letting participants know or not about the upcoming landing condition prior to the execution of the movement. Throughout the training, the magnitude of the perturbation was progressively increased and uncertainty about the task was introduced.

Both the PPT and the CON training took about 20–40 min to perform depending on the training progression. Intensities for the training sessions for the PPT were adjusted to match the intensities of the respective session in the CON group according to the perceived physical demand feedback we received while designing the PPT prior to the study. Both training types followed a progressive design, with increasing intensity and duration toward the latter sessions. Both the experimental and the control group were given instruction and illustration about the exercises prior to the first training session and were supervised by athletic coaches.

Test procedure

The test protocol consisted of two athletic tasks, which were demonstrated to the participants and practiced before testing. The LRJ task was used as most of the motion control is concentrated in the frontal plane, and it has been suggested that greater joint excursions of the trunk, hip and knee in the frontal plane increase the risk of ACL injury (Hewett et al., 2005b; Hewett & Myer, 2011). Participants started in an upright standing position and performed LRJ by jumping over a predefined distance with the right foot to the left side

Table 1. Overview of the plyometric training (CON) progression

Session (CON)	Exercises*
1 & 2	Wall touches (3 × 30 s) Split squat jumps (2 × 40) Tuck jumps (2 × 30) Cone hops with 180° turn (4 cones × 10)
3 & 4	Wall touches (4 × 30 s) Split squat jumps (2 × 50) Tuck jumps (2 × 40) Cone hops with 180° turn (4 cones × 20)
5 & 6	Wall touches (5 × 30 s) Split squat jumps (2 × 60) Tuck jumps (2 × 50) Cone hops with 180° turn (4 cones × 30)
7 & 8	Wall touches (5 × 30 s) Split squat jumps (2 × 60) Tuck jumps (2 × 50) Cone hops with 180° turn (4 cones × 30)
9 & 10	Drop jumps: 45 cm (20) Wall touches (5 × 30 s) Split squat jumps (2 × 60) Tuck jumps (2 × 50) Cone hops with 180° turn (4 cones × 30)
11 & 12	Drop jump: 45 cm (30) Wall touches (6 × 30 s) Split squat jumps (2 × 70) Tuck jumps (2 × 60) Cone hops with 180° turn (4 cones × 40) Drop jump 45 cm (40)

*30 s break between sets and 2 min between exercises.

before pushing off with their left leg to get back to their starting position (Celebrini et al., 2014; Mornieux et al., 2014b; Weltin et al., 2015). The jumping distance was calculated for every subject individually with respect to individual leg length to ensure comparable load configuration of the lower extremity muscles. This resulted in jump distances ranging from 1.00 to 1.26 m. Subjects were instructed to keep looking forward while performing the maneuver as fast as possible. Additionally, cutting maneuvers were tested to mimic an athletic high-risk task associated with knee injury (Olsen et al., 2004; Alentorn-Geli et al., 2009b). To create an unanticipated set-up, a three-light guiding system was used to indicate which of the following tasks should be executed: a straight run, a cutting maneuver 45° from the direction of travel (cutting to the right with the left foot planted on the force platform) and a cross-cut 20° from the direction of travel. The approach speed was set to 4.0 ± 0.2 m/s and controlled by two timing gates, and the light cue occurred approximately 650 ms before the left foot contacted the force plate. Trials were not used for final data analysis if subjects did not match the specified jumping distance or cutting angle, could not get back into the original position, or looked down at the landing place during the maneuver. Kinematics and GRFs of one representative subject are presented for the LRJ (Fig. 2a) as well as the cutting maneuver (Fig. 2b). All participants underwent post-tests immediately after finishing the 4-week training period.

Data collection

The different tasks were performed on a force plate (sampling frequency 2000 Hz, OR6-6-2000; Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA). Three-dimensional whole-body kinematics were recorded using 20 reflective markers placed on anatomical landmarks on the

Table 2. Overview of the perturbation-enhanced plyometric training (PPT) progression

Session (PPT)	Exercises*
1	Lateral reactive jumps (2 × 30; each leg) Lateral reactive jumps on the motor-driven plate 7 cm sliding (2 × 30; each leg) Lateral reactive jumps on the motor-driven plate 7 cm counteracting (2 × 30; each leg)
2	Lateral reactive jumps (2 × 30; each leg) Lateral reactive jumps on the motor-driven plate 10 cm sliding (2 × 30; each leg) Lateral reactive jumps on the motor-driven plate 10 cm counteracting (2 × 30; each leg)
3	Lateral reactive jumps (2 × 35; each leg) Lateral reactive jumps on the motor-driven plate 12 cm sliding (2 × 35; each leg) Lateral reactive jumps on the motor-driven plate 12 cm counteracting (2 × 35; each leg)
4	Lateral reactive jumps (2 × 35; each leg) Lateral reactive jumps on the motor-driven plate random [7 cm sliding or stable] (2 × 35; each leg) Lateral reactive jumps on the motor-driven plate random [7 cm counteracting or stable] (2 × 35; each leg)
5	Lateral reactive jumps (3 × 30; each leg) Lateral reactive jumps on the motor-driven plate random [10 cm sliding or stable] (2 × 35; each leg) Lateral reactive jumps on the motor-driven plate random [10 cm counteracting or stable] (2 × 35; each leg)
6	Lateral reactive jumps (3 × 30; each leg) Lateral reactive jumps on the motor-driven plate random [12 cm sliding or stable] (2 × 35; each leg) Lateral reactive jumps on the motor-driven plate random [12 cm counteracting or stable] (2 × 35; each leg)
7 & 8	Lateral reactive jumps (3 × 35; each leg) Lateral reactive jumps on the motor-driven plate random [7 cm sliding/counteracting or stable] (4 × 40; each leg)
9 & 10	Lateral reactive jumps (3 × 40; each leg) Lateral reactive jumps on the motor-driven plate random [10 cm sliding/counteracting or stable] (4 × 40; each leg)
11 & 12	Lateral reactive jumps (3 × 40; each leg) Lateral reactive jumps on the motor-driven plate random [12 cm sliding/counteracting or stable] (4 × 40; each leg)

*30 s break between sets and 2 min between exercises.

trunk, pelvis, and right leg via self-adhesive, double-sided tape. The markers (Ø14 mm) were placed on the suprasternal notch, the transition between the body of the sternum and the xiphoid process, the T6 vertebra, the anterior and posterior superior iliac spines, the great trochanter, medial and lateral epicondyles of the knee, the tibia (two markers), the medial and lateral malleoli, the first and fifth metatarsal head, the second proximal phalange, and the posterior, medial, and lateral aspects of the heel cup, respectively. This marker placement was used in previously published articles for the lower limb (Gehring et al., 2009) and the trunk (Mornieux et al., 2014a; Weltin et al., 2015, 2016). A 12-camera motion analysis system (Vicon V-MX, VICON Motion System Ltd., Oxford, UK) was used to record marker trajectories at a sampling

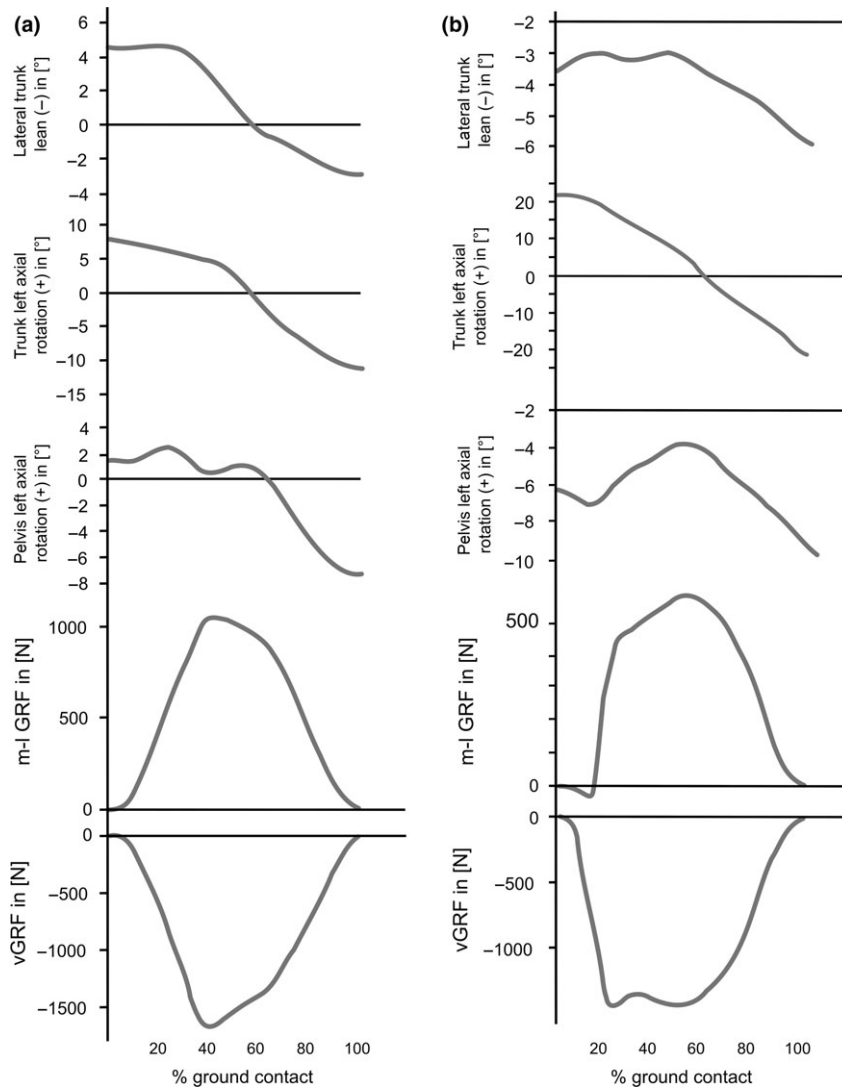


Fig. 2. Main trunk and pelvis kinematic as well as medio-lateral and vertical GRFs data of one representative participant performing a LRJ trial (a) and an unanticipated cutting maneuver (b).

frequency of 200 Hz. A static trial was recorded with the subject standing in a predefined neutral position to characterize the underlying parameters of the kinematical model. The model consists of six segments (forefoot, calcaneus, shank and thigh of the left leg, as well as the pelvis and trunk). Segment length and joint centers were calculated during the static trial for further motion analysis (Grood & Suntay, 1983; Wu et al., 2002). Joint kinematics was calculated using a yxz Euler rotation sequence of the respective segment coordinate systems and expressed with reference to the static trial. A 3D inverse dynamic model was used to calculate the external joint moments starting at the most distal joint (Vicon Nexus 1.6.1; VICON Motion Systems Ltd., Oxford, UK).

Data analysis

All kinematic and kinetic data were low pass filtered at 14 Hz (4th order, zero-lag Butterworth) based on a residual analysis. Initial contact (IC) was defined when the vGRF first transgressed 10N. Knee kinematics, pelvis, and trunk kinematics as measures of core control, as well as knee kinetics, were analyzed at the time of peak knee abduction moment (external

moment) within the first 20% of the foot contact phase, because most non-contact ACL injuries occur during this phase (Olsen et al., 2004). Step width, defined as the distance between foot and mid-pelvis, was analyzed at IC, to reflect possible alterations in cutting technique (Dempsey et al., 2009). Knee kinetics were analyzed in all three planes of motion, while trunk and pelvis kinematics were analyzed only in the frontal and transverse plane, because (a) multi-planar knee joint loading elicits the most strain on the ACL (Markolf et al., 1995; Oh et al., 2012) and (b) because of the influence of the frontal and transverse plane kinematics on knee joint load for trunk and pelvis (Frank et al., 2013; Weltin et al., 2015). All joint angles refer to intersegment angles, describing the motion of the distal segment relative to the proximal segment, except for trunk and pelvis excursion, which are expressed relative to the world axis.

Statistical analysis

For each subject, six successful trials of LRJ and unanticipated cuts were averaged and served as the basis for the statistical analysis, performed using SPSS (SPSS 20.0, Inc.,

Chicago, Illinois, USA). Kolmogorov–Smirnov tests revealed the normal distribution of all acquired data. An analysis of variance (ANOVA) with two factors (group × time) was performed on all the dependent variables for both tasks separately, with group (PPT vs CON) as a between-subject factor and time (pre vs post) as a repeated within-subject factor. A Bonferroni-corrected paired comparison design was used to further analyze the impact of the group when a statistically significant interaction effect for time and group was observed. The alpha level was set to 0.05. All results are expressed as group means ± SD including partial η^2 values to estimate effect sizes. Effect sizes were considered small ($0.01 < \eta^2 < 0.06$), medium ($0.06 < \eta^2 < 0.14$) and large ($\eta^2 > 0.14$) according to Cohen (1988).

Results

LRJ (Table 3)

We found significant main effects of time for trunk ($P < 0.001$, $\eta^2 = 0.596$) and pelvis rotations ($P = 0.003$, $\eta^2 = 0.337$) as well as knee flexion ($P = 0.028$, $\eta^2 = 0.200$) and knee rotation moments ($P < 0.001$, $\eta^2 = 0.472$). Additionally, there were significant interaction effects of time and group for trunk ($P = 0.003$, $\eta^2 = 0.339$) and pelvis rotations ($P = 0.048$, $\eta^2 = 0.116$). Specifically, PPT demonstrated reduced trunk rotation away from the new movement direction (Fig. 3, $P < 0.001$, $\eta^2 = 0.799$) and increased pelvis rotation in the new movement direction ($P = 0.006$, $\eta^2 = 0.511$) after training, while decreased extension and internal rotation moments were true for both trainings.

Cuts (Table 4)

Significant main effects of time were found for trunk ($P = 0.008$, $\eta^2 = 0.277$) and pelvis rotations ($P = 0.049$, $\eta^2 = 0.165$) as well as step width ($P = 0.029$, $\eta^2 = 0.199$). In addition, we found a significant interaction effect of time and group for step width

($P = 0.025$, $\eta^2 = 0.208$). To be precise, after both trainings, trunk rotation away from the new movement direction was reduced, while pelvis rotation toward the new movement direction was increased. The analysis of the impact of the training within the specific groups showed decreased step width ($P = 0.002$, $\eta^2 = 0.577$) after PPT training. Albeit no significant interaction effect were reported for trunk rotation, it is worth noticing that PPT showed a mean reduction of 7.2° for trunk rotation against the new movement direction compared to a mean reduction of 3.9° for CON.

Discussion

The main findings of this study were (a) that during LRJs, PPT effectively reduced trunk rotation away from the new direction of travel and increased pelvis rotation in the new direction of travel, while both trainings established better knee loading patterns during LRJs, and (b) that during unanticipated cuts, step width was reduced by perturbation-enhanced plyometric training, while both trainings reduced trunk rotation against the new direction of travel and increased pelvis rotation toward the change of direction. Our hypothesis that the perturbation training would be effective in changing lower limb biomechanics as well as trunk and pelvis control was partly supported for LRJs.

The hypothesis that perturbation-enhanced plyometric training would change trunk control in the frontal plane during lateral movements must be rejected. Despite the application of LRJ in the PPT, where most of the whole-body movement control is centered in the frontal plane, we did not observe any changes in lateral trunk lean. However, recent studies investigating the effect of trunk stabilization training (Jamison et al., 2012b) and a core stability program

Table 3. Overview of all tested parameters for the LRJs. Values represents the grand mean of all subjects and all trials within the specific training groups. Alpha level was set at 0.05 and in case of a significant time × group interaction effect, significant pre- to post-changes are bold and denoted with*

	Intervention group	Pre-test	Post-test	Time, <i>P</i> -values (pooled groups)	Time × group, <i>P</i> -values	<i>P</i> -value (pre-post)
Lateral trunk lean (–) [°]	CON	–0.5 ± 5.5	–1.2 ± 4.1	0.117	0.637	–
	PPT	4.6 ± 1.9	3.3 ± 2.4			
Trunk left axial rotation [°]	CON	13.4 ± 10.0	10.0 ± 10.0	0.000	0.003	0.425
	PPT	13.8 ± 6.6	0.4 ± 7.2*			0.000
Medial pelvic tilt [°]	CON	1.8 ± 3.4	1.4 ± 3.7	0.075	0.357	–
	PPT	3.7 ± 2.6	2.4 ± 2.3			
Pelvis left axial rotation (+) [°]	CON	0.7 ± 7.2	–0.7 ± 8.3	0.003	0.048	0.315
	PPT	2.5 ± 3.9	–3.6 ± 4.0*			0.006
Knee flexion moment (–) [Nm/kg]	CON	0.14 (–0.073 to 0.35)	0.03 (–0.09 to 0.16)	0.028	0.680	–
	PPT	0.14 (0.03–0.21)	–0.02 (–0.14 to 0.11)			–
Knee abduction moment [Nm/kg]	CON	0.31 (0.24–0.37)	0.3 (0.24–0.36)	0.141	0.073	–
	PPT	0.22 (0.17–0.28)	0.29 (0.22–0.36)			–
Knee internal rotation moment (+) [Nm/kg]	CON	0.09 (0.01–0.16)	–0.03 (–0.09 to 0.03)	0.000	0.818	–
	PPT	0.06 (–0.01 to 0.13)	–0.04 (–0.12 to 0.03)			–

(Pfle et al., 2013) on trunk control did not report any improvements either. Among the existing training paradigms only technique modification training successfully changed trunk positioning, especially during planned cutting maneuvers (Dempsey et al., 2009) underlying the difficulty to train lateral trunk lean. Although it has been suggested that a dynamic approach is more effective for altering trunk control, it seems difficult to influence lateral trunk flexion after a

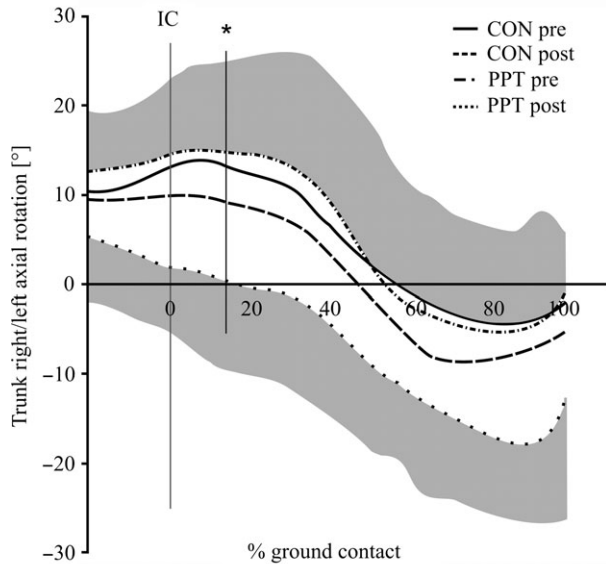


Fig. 3. Grand mean of the trunk rotation for the LRJs from 100 ms before ground contact throughout the whole stance phase. Baseline values are represented by the solid (CON) and the slash dot line (PPT), while post-intervention values are pictured as slashed line (CON) and dotted line (PPT). The gray area represents the standard deviation for all participants. Alpha level was set at 0.05 and * marks significance.

4- to 6-week specific training program. Another likely explanation for that outcome could be the specific set of participants in our study, because the female athletes participating in the tests had an average team sport experience of 10 years with no history of ACL injury during that time. Thus, it might be possible that the participants, if looking at mean values, would not show trunk excursions associated with ACL injury. Indeed, the participants in the perturbation group had already placed the trunk in the new movement direction during LRJ at the baseline measurements and did not change that pattern afterwards. However, it is worth noting that even within this particular set of participants, two females injured their knee during the period when the study was conducted.

With regard to further core control, we observed a reduction in trunk rotation against the new movement direction to an almost neutral position at the point of maximum knee abduction during LRJs after perturbation-enhanced plyometric training and a reduced trunk rotation during unanticipated cuts for both groups, with a large reduction of 7.2° after PPT. It has been shown previously that increased trunk rotation away from the new movement direction was associated with increased knee abduction moments during cuttings (Frank et al., 2013) and unanticipated cuttings (Jamison et al., 2012a). Despite the reduction in trunk rotation, we did not observe any changes in knee abduction loading in either task. There are a variety of reasons why we could not see these changes. One explanation could be that the intervention was not strong enough to elicit changes in knee abduction moment or that the reduction of thorax rotation was not high enough. Another likely explanation could be that the high inter-subject variability of the knee abduction moment

Table 4. Overview of all tested parameters for the unanticipated cutting maneuvers. Values represent the grand mean of all subjects and all trials within the specific training groups. Alpha level was set at 0.05 and in case of a significant time × group interaction effect, significant pre- to post-changes are bold and denoted with*

	Intervention group	Pre-test	Post-test	Time, <i>P</i> -value (pooled groups)	Time × group, <i>P</i> -values	<i>P</i> -value (pre-post)
Lateral trunk lean (-) [°]	CON	-5.6 ± 3.6	-6.2 ± 3.9	0.990	0.278	-
	PPT	-6.5 ± 3.3	-5.6 ± 3.6			
Trunk left axial rotation (+) [°]	CON	14.4 ± 8.9	10.5 ± 8.1	0.008	0.596	-
	PPT	13.6 ± 8.0	6.4 ± 8.8			
Medial pelvic tilt [°]	CON	13.0 ± 6.0	13.2 ± 5.9	0.676	0.478	-
	PPT	12.9 ± 5.7	11.9 ± 5.7			
Pelvis right axial rotation (-) [°]	CON	-7.8 ± 5.8	-9.2 ± 5.7	0.049	0.538	-
	PPT	-10.8 ± 5.5	-13.3 ± 5.6			
Knee flexion moment [Nm/kg]	CON	0.44 (-0.45 to 1.34)	0.03 (-0.09 to 0.16)	0.898	0.174	-
	PPT	0.23 (-0.35 to 0.84)	-0.17 (-0.48 to 0.15)			
Knee abduction moment [Nm/kg]	CON	0.38 (0.23-0.53)	0.52 (0.17-0.88)	0.605	0.263	-
	PPT	0.39 (0.14-0.64)	0.34 (0.16-0.52)			
Knee internal rotation moment (+) [Nm/kg]	CON	0.10 (0.03-0.16)	0.12 (0.05-0.20)	0.424	0.716	-
	PPT	0.00 (-0.22 to 0.21)	0.06 (-0.01 to 0.12)			
Step width [m]	CON	0.41 ± 0.07	0.41 ± 0.06	0.029	0.025	0.969
	PPT	0.43 ± 0.04	0.39 ± 0.05*			0.003

in both groups concealed possible changes due to the intervention. In addition, we were able to report increased pelvic rotation in the new direction of travel during LRJ after PPT, which is considered to be a beneficial strategy for direction changes (Dempsey et al., 2009; Celebrini et al., 2014). This adaptation, however, was not specific to PPT during cutting maneuvers. These changes reflect an improvement of core control during dynamic athletic tasks.

Although the knee joint abduction moment was not influenced by any of the training programs, the knee loading pattern during LRJ after training showed a slight change from knee extension moment to knee flexion moment and a switch from knee internal rotation moment to knee external rotation moment. In particular, the change of the knee rotation moment is in line with the results of Dempsey et al. (2007), who reported a positive association between trunk rotation and peak knee internal rotation moment. A knee loading pattern of combined knee internal rotation moment and knee extension moment together with knee abduction moment is argued to be the most dangerous for the ACL (Markolf et al., 1995; Oh et al., 2012). Oh et al. (2012) even concluded that the internal rotation loading is what strains the ACL the most. Thus, these changes in the knee loading pattern after plyometric and PPT trainings could be assumed to be beneficial with regard to ACL strain.

Finally, PPT participants showed a decrease in step width while no changes occurred for the control group. We found mean decreases in step width of 4 cm after training, which is line with the findings of a study by Dempsey et al. (2009), focusing on the change of the cutting technique. Dempsey et al. (2009) concluded that cutting technique modifications including reduced step width effectively decrease knee joint loading and therefore might reduce the risk of ACL injuries.

Overall, the present results support the concept of training specificity. The influence of perturbation-enhanced plyometric training on core control and to a less extent on knee loading occurred primarily during LRJ, as this testing condition was very close to the training design. A generalization of these changes in core control during unanticipated cuts was limited, as this new training did not further improve core control significantly compared to the plyometric training, despite a large reduction of 7.2° in trunk rotation after PPT.

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A possible limitation of this intervention could be that it did not influence lateral trunk lean, one of the main parameters targeted by PPT. This could be a result of the intervention design incorporating only 12 training sessions over 4 weeks, which may not be enough to elicit changes in lateral trunk lean by healthy and athletic women. Nonetheless, the intervention did change knee loading patterns and especially improved trunk and pelvis control in the transverse plane during LRJs, which indicates that 12 sessions over 4 weeks were sufficient to induce changes. Another factor to consider is that possible changes elicited by the intervention might be concealed by the improvement of the control group as well, which stems from the neuromuscular training that was performed by the control group to ensure comparable training loads between groups. However, this approach ensured to attribute observed changes to the specificity of perturbation-enhanced training.

Perspectives

In summary, a 4-week training program based on perturbation-enhanced plyometric exercises improved core control by reducing trunk rotation away from the cutting direction and established an improved knee loading pattern, possibly reducing the risk of ACL injuries. In the future, all athletes could use PPT worldwide by applying the principles of PPT to materials readily available at the gym. For instance, elastic ropes could be used to create resistance while performing LRJ, and unstable surfaces such as foam pads could create unstable landing conditions. Additionally, these results strongly suggest that trunk stabilization training using dynamic movements is necessary to elicit improvements in core control and lower limb biomechanics. Further research is needed to elucidate whether ACL injury rates would indeed decrease in response to perturbation-enhanced plyometric training.

Key words: ACL injury, trunk control, knee joint control, neuromuscular training.

Supporting Information

Additional Supporting Information may be found in the online version of this article.

Video S1. Example of the landing modalities during the PPT.

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ORIGINAL ARTICLE

Effect of gender on trunk and pelvis control during lateral movements with perturbed landing

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Abstract

In lateral reactive movements, core stability may influence knee and hip joint kinematics and kinetics. Insufficient core stabilisation is discussed as a major risk factor for anterior cruciate ligament (ACL) injuries. Due to the higher probability of ACL injuries in women, this study concentrates on how gender influences trunk, pelvis and leg kinematics during lateral reactive jumps (LRJs). Perturbations were investigated in 12 men and 12 women performing LRJs under three different landing conditions: a movable landing platform was programmed to slide, resist or counteract upon landing. Potential group effects on three-dimensional trunk, pelvic, hip and knee kinematics were analysed for initial contact (IC) and the time of peak pelvic medial tilt (PPT). Regardless of landing conditions, the joint excursions in the entire lower limb joints were gender-specific. Women exhibited higher trunk left axial rotation at PPT (women: $4.0 \pm 7.5^\circ$, men: $-3.1 \pm 8.2^\circ$; $p = 0.011$) and higher hip external rotation at both IC and PPT ($p < 0.01$). But women demonstrated higher knee abduction compared to men. Men demonstrated more medial pelvic tilt at IC and especially PPT (men: $-5.8 \pm 4.9^\circ$, women: $0.3 \pm 6.3^\circ$; $p = 0.015$). Strategies for maintaining trunk, pelvis and lower limb alignment during lateral reactive movements were gender-specific; the trunk and hip rotations displayed by the women were associated with the higher knee abduction amplitudes and therefore might reflect a movement strategy which is associated with higher injury risk. However, training interventions are needed to fully understand how gender-specific core stability strategies are related to performance and knee injury.

Keywords: *Trunk, movement strategy, knee joint, ACL injury*

Introduction

Most anterior cruciate ligament (ACL) injuries occur in non-contact situations, usually direction changes such as cutting or pivot turning, and result in lower limb alignment characterised by knee valgus, together with an internally or externally rotated tibia and an almost fully extended knee (Alentorn-Geli et al., 2009). Most research on the mechanisms of ACL injury is based on knee joint biomechanics. Prospective studies have explored the association between hip muscle weakness and knee injury (Leetun, Ireland, Willson, Ballantyne, & Davis, 2004; Nadler, Malanga, Deprince, Stitik, & Feinberg, 2000). It has also been reported that hip adduction is the primary source of excessive knee valgus (Hollman et al., 2009). Several studies have investigated the influence

of trunk position on lower limb mechanics during landings (Blackburn & Padua, 2008) and/or cutting manoeuvres (Frank et al., 2013; Jamison, Pan, & Chaudhari, 2012). Studies of gender differences in landing and cutting movements reported more knee abduction and hip adduction as well as less internal knee rotation and more external hip rotation in women than men (Beaulieu, Lamontagne, & Xu, 2009; McLean, Lipfert, & Van Den Bogert, 2004). Unfortunately, these studies did not investigate trunk motion. Interestingly, studies of real injury situations revealed that, regardless of gender, injury situations are closely associated with landings in which the trunk is in a noticeably upright position (Sheehan, Sipprell, & Boden, 2012). Furthermore, prospective studies (Zazulak, Hewett, Reeves, Goldberg, & Cholewicki, 2007a, 2007b) found evidence that deficits in trunk

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neuromuscular control and proprioception are strong predictors of ACL injuries in female athletes but not male athletes. Finally, analysis of the situations in which injuries occurred during basketball showed that female athletes had greater lateral trunk lean than men (Hewett, Torg, & Boden, 2009).

Taken together, these findings suggest that trunk and hip joint control might be gender-specific and – when compromised – a potential risk factor for knee joint injuries. A better understanding of the relationship between gender-specific core stability strategies and the biomechanics of ACL injury mechanisms is necessary to improve the design of ACL injury prevention programmes.

Core stability has been defined as the ability to control the position and the motion of the trunk over the pelvis to allow controlled, optimal production and transfer of force and motion to the proximal segments during athletic activities (Kibler, Press, & Sciascia, 2006). While the “core” includes all anatomical body segments between the shoulders and the knee (Hibbs, Thompson, French, Wrigley, & Spears, 2008), the trunk and pelvis are probably the most important to core stability. It has been shown that poor lateral trunk control, measured by trunk excursions after perturbation generated by a sudden release of force, is associated with a higher risk of knee injury. The pelvis is the functional link between the trunk and the lower extremities. Effective control of the pelvis, especially in the frontal plane, is essential for optimal alignment of the axis of the lower limbs during changes of direction (Houck, Duncan, & De Haven, 2006). Core stability during lateral movements may therefore be described as the ability to control trunk and pelvis motion to ensure that the axis of the lower limbs is properly aligned, even if perturbation occurs during the movement execution. These findings suggest that research into gender differences in core stability during lateral movements should investigate trunk and pelvis motion jointly with lower limb motion and include the use of perturbation during movement tasks.

The aim of this study was to determine how gender influenced trunk, pelvis and leg kinematics during lateral reactive jumps (LRJs) with perturbed landing conditions. More specifically, we hypothesised that female athletes would demonstrate higher trunk excursions and different pelvis and hip biomechanics in the frontal and transversal plane, as well as greater knee abduction angles.

Methods

Subjects

A total of 12 women (23.2 ± 2.6 years, 1.67 ± 0.05 m, 60.7 ± 7.3 kg with 10.6 ± 2.9 years experience of

team sports) and 12 men (23.2 ± 3.0 years, 1.80 ± 0.06 cm, 76.4 ± 6.8 kg, 12.9 ± 3.1 years experience of team sports) with no history of (ACL) injury or neurological disorders participated in this study. All participants were informed about the purpose, procedures and possible risks associated with the experimental set-up and gave their written consent prior to testing. The experiments were run in accordance with the Declaration of Helsinki and were approved by the Ethics Committee of the Albert-Ludwigs University of Freiburg.

Protocol

The LRJ task was used because in this task motion control is concentrated in the frontal plane, and greater joint excursions of the trunk, hip and knee in the frontal plane are thought to increase the risk of ACL injury (Hewett & Myer, 2011; Hewett et al., 2005). Participants performed LRJs from an upright standing position by jumping a pre-defined distance with the left foot to the right side before pushing off with their right leg to get back to their starting position (Fleischmann, Gehring, Mornieux, & Gollhofer, 2010; Mornieux, Gehring, Tokuno, Gollhofer, & Taube, 2014).

Jump distance was calculated individually on the basis of leg length to ensure a comparable load configuration across the lower limb muscles. This resulted in jump distances ranging from 1.00 m to 1.26 m. Participants were instructed to try to keep looking ahead whilst performing the manoeuvre. The task was performed on a motor-driven plate, which could move linearly to produce the required perturbation conditions. Participants experienced three LRJ conditions: (1) no perturbation (stable condition), (2) rightward perturbation (10 cm slide) and (3) leftward perturbation (counteracting motion over 10 cm; Figure 1).

The three landing conditions were presented in random order in blocks of 10 jumps per condition, matched across groups. Participants knew which of the three LRJ conditions would occur prior to movement execution. Participants were familiarised with the LRJ task before measurements began. Measurements were taken during eight successful LRJ trials in each landing condition. Trials were not used for final data analysis if participants did not achieve the specified jumping distance, failed to get back to their original position or looked down at the landing place during the manoeuvre.

Instrumentation

The custom-made motor-driven platforms (Department of Sports Sciences, University of Freiburg, Germany) used in this study are made up of two

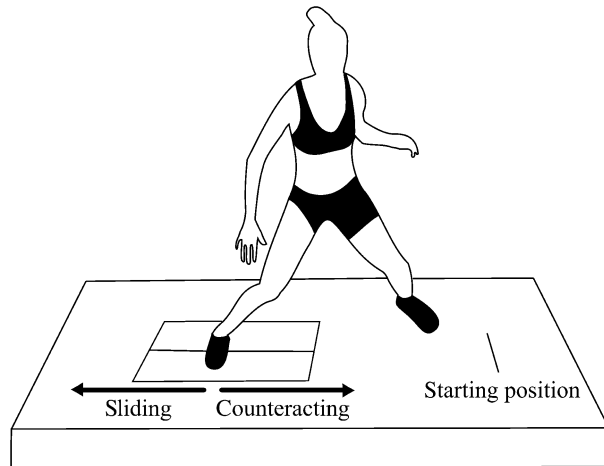


Figure 1. A schematic view of the LRJ task. The landing surface was programmed to slide, resist or counteract upon landing.

independently moving plates (1.15 m × 0.30 m), each driven by a motor. A detailed description of the device has been published elsewhere (Mornieux, Gehring, Tokuno, et al., 2014). Perturbation was triggered by breaking a light beam (M18 series, Panasonic Electric Works Europe AG, Holzkirchen, Germany), 0.04 m above the plate, to compensate for the delay in activation of the motor-driven plate.

Three-dimensional whole body kinematics were recorded using 18 reflective markers ($\varnothing 14\text{mm}$) placed on anatomical landmarks on the trunk, pelvis and the right leg. Markers were placed on the suprasternal notch, the transition between the body of the sternum and the xiphoid process, the T6 vertebra, the anterior and posterior superior iliac spines, the great trochanter, the medial and lateral epicondyles of the knee, the tibia (two markers), the medial and lateral malleoli, the first metatarsal head, and the posterior, medial and lateral aspects of the heel cup. This marker placement was based on previously published research on lower limb (Gehring, Melnyk, & Gollhofer, 2009) and trunk (Mornieux, Gehring, Fürst, & Gollhofer, 2014) movements. Marker trajectories were recorded with a 12-camera motion analysis system (Vicon V-MX, VICON Motion System Ltd., Oxford, UK) at a sampling frequency of 200 Hz. To define the underlying parameters of the five-segment kinematical model (calcaneus, shank and thigh of the left leg, pelvis and trunk), a static trial was recorded with the participant standing in a pre-defined neutral position. During the static trial segment, length and joint centres were calculated (Vicon Nexus 1.6.1, VICON Motion Systems Ltd., Oxford, UK) for use in subsequent motion analysis (Grood & Suntay, 1983; Wu et al., 2002). Joint kinematics were calculated using a YXZ Euler rotation sequence of the respective segment coordinate systems and expressed with reference to the static trial. In anatomical terms,

the rotation sequence started with a rotation around the flexion/extension axis, followed by a rotation around the adduction/abduction axis and with a rotation around the internal/external axis at the end.

Data analysis

All kinematic data were low-pass filtered at 14 Hz based on a residual analysis (Winter, 2005). Initial contact (IC) and toe-off were determined from the first metatarsal marker's vertical acceleration signal (Maiwald, Sterzing, Mayer, & Milani, 2009). Foot, knee, hip, pelvis and trunk kinematics were analysed at IC to determine the movement strategies used in the various LRJ conditions. Kinematic variables were also analysed at the time of peak pelvis medial tilt (PPT) angle during the first 50% of the foot contact phase, to investigate the consequences of the various movement strategies used during the weight acceptance phase. The PPT (Figure 2C) was chosen because the pelvic structure is the central element of the core, and pelvis tilt is an important factor in lower limb alignment during weight acceptance (Houck et al., 2006). Medial pelvic tilt was a tilt in the new movement direction (Houck et al., 2006). Differences between landing conditions at PPT should mainly reflect the consequences of mechanical perturbation, rather than differences in the trunk and pelvis control strategies used by the participants, and so analysis of the influence of the landing condition on trunk and pelvis control strategies was based primarily on measurements at IC.

All variables except foot kinematics were analysed in all three planes of motion because (1) the frontal plane is relevant to the mechanisms of ACL injury (Hewett & Myer, 2011); (2) the influence of the sagittal plane on ACL injury mechanism in the case of the hip, knee and trunk; and (3) the influence of transverse plane kinematics on knee joint load for the hip, knee and trunk (Alentorn-Geli et al., 2009). Foot kinematics to assess the foot progression angle (FPA) were only calculated in the transversal plane (Fleischmann et al., 2010). All joint angles were expressed as intersegment angles except trunk and pelvis excursion which were expressed relative to the world axis. The reliability values (intraclass correlation coefficients, ICCs) were all above 0.86 for all variables for the LRJs. The standard error of measurement (SEM) was below 1.9° for all parameters analysed. The specific values for some selected parameters are as follows (mean across conditions): lateral trunk lean (ICC: 0.913; SEM: 0.7°), axial trunk rotation (ICC: 0.891; SEM: 1.1°), pelvis medial tilt (ICC: 0.970; SEM: 0.9°), axial pelvis rotation (ICC: 0.955; SEM: 1.1°), hip external rotation (ICC: 0.985; SEM: 1.4°) and knee abduction (ICC: 0.990; SEM: 1.0°).

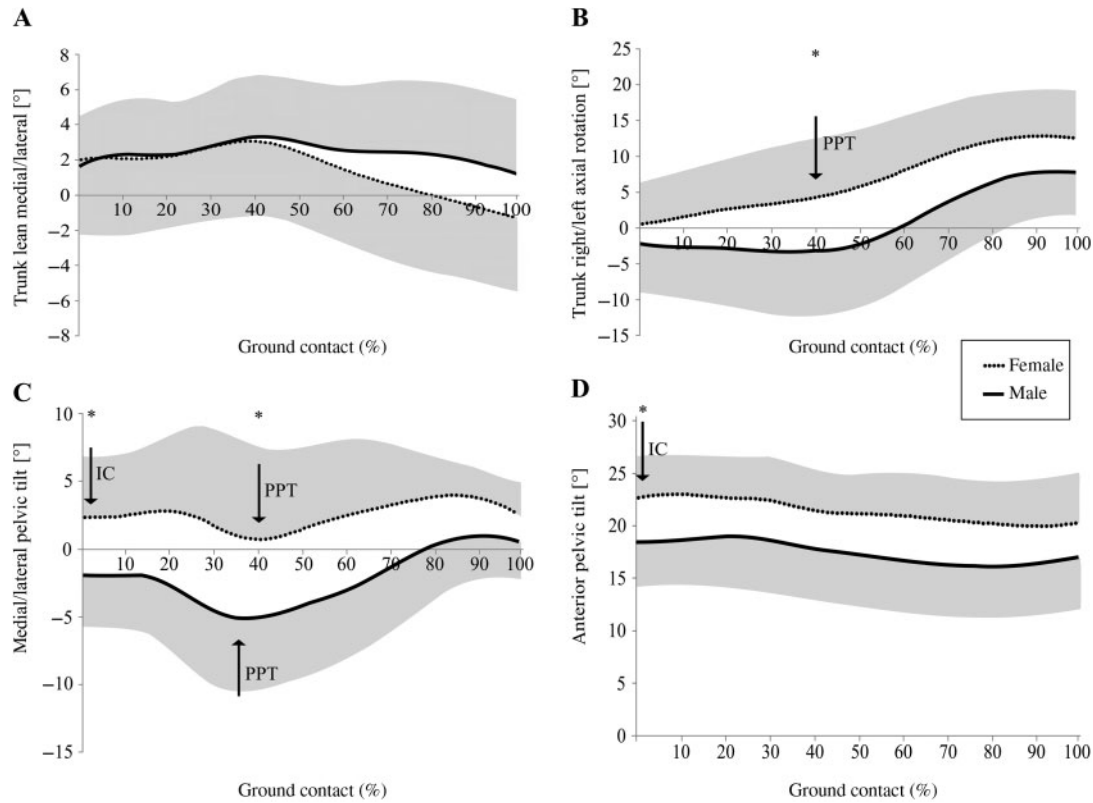


Figure 2. Trunk angles in the frontal plane (A) and transversal plane (B) and pelvis angles in the frontal plane (C) and sagittal plane (D), for men (solid line) and women (dotted line). Curves represent group means, the error bars represent standard deviation. The alpha level was set to 0.05; *denotes a significant difference between men and women.

Statistical analysis

Eight successful trials per participant in each landing condition were averaged to provide data for statistical analysis. Statistical analysis was performed using SPSS version 20.0 (SPSS, Inc., Chicago, IL). Kolmogorov–Smirnov tests indicated that all the variables were normally distributed. An analysis of variance with two factors (gender and landing condition) was performed on all the dependent variables, with gender (male and female) as a between-subjects factor and landing condition (stable, sliding and counteracting) as a repeated within-subject factor. Pearson's correlation coefficients were calculated for the correlations between the trunk, pelvis, hip and knee. Main effects were analysed further using Bonferroni corrected paired comparisons. The alpha level was set to 0.05. All results are expressed as group means \pm SD.

Results

Gender

Compared with men, women exhibited significant greater left axial trunk rotation at PPT (women: $4.0 \pm 7.5^\circ$, men: $-3.1 \pm 8.2^\circ$; $F = 5.0$, $p = 0.036$; Figure 2B) but not at IC. Women also had greater external

hip rotation at IC (women: $-5.6 \pm 6.2^\circ$, men: $2.2 \pm 5.6^\circ$; $F = 10.6$, $p = 0.004$; Figure 3A) and PPT (women: $-3.0 \pm 4.0^\circ$, men: $4.4 \pm 5.2^\circ$; $F = 15.3$, $p = 0.001$, Figure 3A). However, in the other planes, joint excursion for hip and trunk was similar at both IC and PPT (Figure 2A). Women had pronounced knee abduction angles, whereas men had adduction angles at IC (women: $-4.6 \pm 3.8^\circ$, men: $1.1 \pm 3.7^\circ$; $F = 13.3$, $p = 0.001$; Figure 3B) and PPT (women: $-5.7 \pm 5.3^\circ$, men: $5.1 \pm 5.0^\circ$; $F = 26.1$, $p = 0.001$; Figure 3B), but there were no gender differences in knee kinematics in the sagittal or transverse planes.

Men demonstrated greater medial pelvic tilt at IC (men: $-2.1 \pm 3.5^\circ$, women: $2.5 \pm 4.2^\circ$; $F = 8.0$, $p = 0.010$; Figure 2C) and PPT (men: $-5.8 \pm 4.9^\circ$, women: $0.3 \pm 6.3^\circ$; $F = 7.0$, $p = 0.015$; Figure 2C). Men also demonstrated less anterior pelvic tilt but only at IC (men: $18.9 \pm 5.5^\circ$, women: $23.0 \pm 4.4^\circ$; $F = 5.1$, $p = 0.012$; Figure 2D). There was no gender difference in FPA.

Left axial trunk rotation was associated with increased external hip rotation angles ($r = 0.36$, $p = 0.002$) and increased knee abduction angles ($r = 0.33$, $p = 0.005$). Greater external hip rotation was associated with increased knee valgus ($r = 0.58$, $p < 0.001$). There was a negative association between

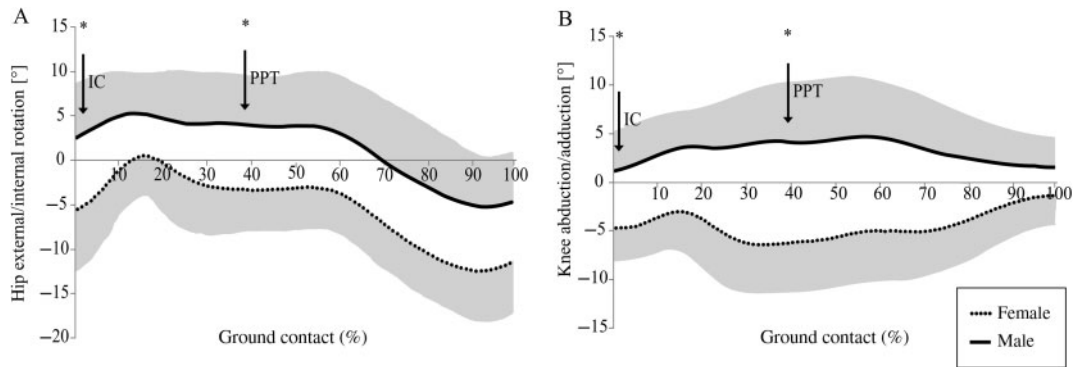


Figure 3. Hip angles in the transversal plane (A) and knee angles in the frontal plane (B), for men (solid line) and women (dotted line). Curves represent group means. The alpha level was set to 0.05; *denotes a significant difference between men and women.

medial pelvic tilt and knee abduction angle ($r = -0.36$, $p = 0.002$).

Landing condition

Analysis revealed that transversal plane, trunk ($F = 5.8$, $p = 0.006$) and pelvis segments ($F = 8.6$, $p = 0.001$), hip joints ($F = 3.8$, $p = 0.015$), knee joints ($F = 9.5$, $p = 0.001$) and FPA ($F = 7.9$, $p = 0.001$) were significantly influenced by landing condition at IC. Landing condition affected sagittal plane kinematics of the hip ($F = 12.5$, $p = 0.001$) and knee joints ($F = 19.1$, $p = 0.001$). In the frontal plane, only pelvic tilt was significantly influenced by landing condition ($F = 8.3$, $p = 0.001$). Overall counteracting led to greater left axial trunk rotation, greater medial tilt and left axial rotation of the pelvis, more external hip and knee rotation and more exorotation of the foot (Table I).

Discussion

The primary finding of this study is that men and women organise LRJs differently at all levels of the kinetic chain (i.e. trunk, pelvis, hip and knee), reflecting a gender difference in core stability strategy. In this study, women used a strategy characterised by greater trunk and hip rotations, whereas men used a strategy characterised by higher medial pelvic tilt angles. Women also executed LRJs with higher knee abduction angles.

At the instant of PPT, female athletes relied on a movement strategy characterised by greater left axial trunk rotation together with greater external hip rotation, meaning that movement control was largely in the transverse plane. Further analysis of the time course of trunk rotation revealed that women had already rotated their trunk to the left at IC and rotated the trunk in the new jumping direction

Table I. Mean \pm SD values at the time of IC for foot, knee, hip, pelvis and trunk parameters compared between the different landing conditions during LRJs

IC		Landing condition (pooled subjects)		
		Stable	Sliding	Counteracting
Trunk	Flexion (+) [°]	15.1 \pm 5.7	15.2 \pm 6.4	15.0 \pm 5.7
	Lateral lean (+) [°]	2.0 \pm 3.9	2.0 \pm 3.5	1.0 \pm 3.1
	Left axial rotation (+) [°]	-2.4 \pm 5.5	-0.9 \pm 5.3	0.2 \pm 5.5^a
Pelvis	Anterior tilt (+) [°]	20.7 \pm 5.0	20.4 \pm 6.2	20.6 \pm 4.8
	Medial tilt (-) [°]	-0.2 \pm 4.3	1.1 \pm 4.6	-0.2 \pm 4.5^{a,b}
	Left axial rotation (+) [°]	1.4 \pm 5.5	1.6 \pm 5.5	3.3 \pm 5.6^{a,c}
Hip	Flexion (+) [°]	34.8 \pm 6.9	36.1 \pm 9.3	33.1 \pm 6.7^{a,c}
	Abduction (-) [°]	-31.0 \pm 5.6	-30.9 \pm 5.1	-31.4 \pm 6.9
	External rotation (-) [°]	-1.8 \pm 7.2	-1.0 \pm 7.5	-2.2 \pm 6.8^{a,b}
Knee	Flexion (+) [°]	27.0 \pm 4.7	29.2 \pm 6.1	25.8 \pm 4.9^{a,b}
	Abduction (-) [°]	-1.8 \pm 5.0	-1.8 \pm 4.5	-1.7 \pm 4.7
	Internal rotation (-) [°]	-3.5 \pm 8.4	-3.1 \pm 8.2	-5.8 \pm 8.5^{a,c}
FPA	Exorotation (-) [°]	-22.3 \pm 7.4	-18.3 \pm 9.0	-22.4 \pm 8.5^{a,b}

Significant main effects between conditions are bolded and denoted with *. Alpha level was set to 0.05.

^aDenotes significant difference from stable.

^bDenotes significance difference from sliding.

^cMarks significance difference from both other conditions.

throughout the whole ground contact phase. Trunk rotation was associated with an increased external hip rotation angle ($r = 0.36$, $p = 0.002$). The higher external hip rotation displayed by women is consistent with findings during unanticipated cuttings (Beaulieu et al., 2009).

Video analysis of ACL injuries has revealed that after reaching a critical knee joint position (position of no return), the hip moves more rapidly into adduction and internal rotation, and the knee collapses in valgus (Ireland, 1999). Lack of strength in the external hip rotators of women (Leetun et al., 2004) may explain why injuries occur in this fashion. From this perspective, the greater external hip rotation displayed by the female athletes in our study may be a strategy for avoiding potentially high risk positions of no return. However, the external hip rotation together with the left axial rotation of the trunk were biomechanically associated with increased knee abduction angles ($r = 0.58$, $p < 0.001$ and $r = 0.33$, $p = 0.005$, respectively) in the present study. Women had knee valgus of -4.6° at IC and -5.7° at PPT, whereas men had varus of 1.1° at IC and 5.1° at PPT. It has previously been reported that female athletes have greater knee abduction angles during unanticipated cuttings (Beaulieu et al., 2009; Ford, Myer, Toms, & Hewett, 2005), indicating that during LRJs female athletes have substantial dynamic knee valgus and possibly higher ACL loads (Markolf et al., 1995). Although the female athletes in our study successfully used left axial trunk rotation and external hip rotation to perform LRJs, this strategy was associated with an increased knee abduction angle, which may increase the probability of overloading knee joint structures. It is likely that control of trunk rotation is important during lateral movements and that during an LRJ performed mainly in the frontal plane participants should restrain trunk rotation.

Male athletes used a strategy based on pelvis control in the frontal plane. More specifically, men had more medial pelvic tilt at IC, and at the moment of PPT, than women. Tilting the pelvis towards the new movement direction has previously been described as a strategy for aligning the lower limb correctly during weight acceptance so as to ensure a proper propulsion phase in cuttings (Houck et al., 2006), which suggests that this pelvic strategy may help men to transfer ground reaction forces to the centre of mass more effectively during LRJs. Additionally, greater medial pelvic tilt implies a reduction of the hip abduction together with a possible reduction of external hip abduction moment. As hip abduction moment is associated with increased knee abduction moment (Frank et al., 2013; Hewett & Myer, 2011), medial pelvic tilt would influence

knee abduction moment. Women however showed lateral pelvic tilt, which was associated with an increased knee abduction angle ($r = 0.36$, $p = 0.002$). This pelvic tilt may be adopted as a means of preventing contralateral hip drop during single leg stance, known as the Trendelenburg sign (Russell, Palmieri, Zinder, & Ingersoll, 2006), and is often associated with a lack of strength in the hip abductors (Leetun et al., 2004), which has been reported to be more common in women than men (Powers, 2010). A further difference was that men showed right axial trunk rotation at IC and at the corresponding PPT. Detailed analysis of the time course of trunk rotation (Figure 2B) revealed that men showed phase-specific organisation of trunk orientation, namely a stable trunk during the weight acceptance phase and an active rotation towards the new direction of travel during propulsion.

Therefore, athletic programmes could target pelvic kinematics to train participants to move with stable trunk orientation; for example, dynamic core stability exercises could be used to improve strength and muscular coordination.

The LRJ landing condition influenced the kinematic parameters on initial impact regardless of gender. Changes at IC reflected the different feedforward motor control strategies. Landing condition influenced the positioning of all joints in the transversal plane. The counteracting condition resulted in more trunk and pelvis left axial rotation, more external hip rotation and more exorotation of the foot. Higher exorotation of the foot was previously reported to be a strategy for shifting loading from the frontal plane to the sagittal so as to reduce knee loading in the frontal plane (Fleischmann et al., 2010). Sagittal plane kinematics of the hip and knee were influenced by landing condition; the leftward perturbation produced lower knee and hip flexion angles at IC, which suggests that a more extended joint position was being used to better compensate the counteracting motion. In the frontal plane, however, only the pelvic segment was influenced by the landing condition and by only a few degrees.

In summary, women performed LRJs using a strategy based on greater left axial trunk rotation and greater external hip rotation, which was associated with higher knee abduction values than observed in their male counterparts. Men used a pelvic-centred strategy that seemed to improve transmission of forces during the push-off phase. Further research is needed to improve the understanding of how gender-specific core stability strategies influence performance parameters and knee injury risk during lateral movements.

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Anticipatory postural adjustments during cutting manoeuvres in football and their consequences for knee injury risk

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Abstract

Anticipatory postural adjustments (APAs), i.e. preparatory positioning of the head, the trunk and the foot, are essential to initiate cutting manoeuvres during football games. The aim of the present study was to determine how APA strategies during cutting manoeuvres are influenced by a reduction of the time available to prepare the movement.

Thirteen football players performed different cutting tasks, with directions of cutting either known prior to the task or indicated by a light signal occurring 850, 600 or 500 ms before ground contact.

With less time available to prepare the cutting manoeuvre, the head was less orientated towards the cutting direction ($P = 0.033$) and the trunk was even more rotated in the opposite direction ($P = 0.002$), while the foot placement was not significantly influenced. Moreover, the induced higher lateral trunk flexion correlated with the increased knee abduction moment ($r = 0.41$; $P = 0.009$).

Increasing lateral trunk flexion is the main strategy used to successfully perform a cutting manoeuvre when less time is available to prepare the movement. However, higher lateral trunk flexion was associated with an increased knee abduction moment and therefore an increased knee injury risk. Reducing lateral trunk flexion during cutting manoeuvres should be part of training programs seeking the optimisation of APAs.

Keywords: *change of direction, trunk, knee abduction moment, step width, head rotation*

Introduction

Changing the direction or avoiding obstacles is an integral part of many team sports as for instance football. It is evident that preparatory strategies, so-called anticipatory postural adjustments (APAs), are necessary for a successful movement execution (Jindrich & Qiao, 2009; Patla, Adkin, & Ballard, 1999; Xu, Carlton, & Rosengren, 2004). Basically, changing direction requires a reorientation of the body together with the impulse vector to move the centre of mass (CoM) in the direction of the intended movement. This could be achieved by three main strategies involving the whole body.

First, the foot placement via the step width regulation plays an important role (Patla et al., 1999; Vallis & McFadyen, 2003; Wheeler & Sayers, 2010). Indeed, the placement of the foot determines the position of the centre of pressure and defines in which direction the CoM can be accelerated (Patla et al., 1999). Second, head orientation is used to

adjust the visual field and to initiate the reorientation of the whole body (Hicheur, Vieilledent, & Berthoz, 2005; Sreenivasa, Frissen, Souman, & Ernst, 2008; Vallis & McFadyen, 2005). Finally, the control of trunk motion is crucial (Sreenivasa et al., 2008; Vallis & McFadyen, 2005). The trunk yaw (rotation about the long axis of the body) orientates the body in the new direction. Moreover, the trunk roll (lateral flexion) is used to move lower limbs and upper body in opposite directions, which allows moving the CoM towards the new direction during the stance phase (Patla et al., 1999), especially when the change of direction occurs late. It is worth noting that APAs must already be planned and initiated in the step before changing the direction (Patla et al., 1999; Vallis & McFadyen, 2005).

Changes of direction can be performed differently, e.g. side-stepping, cutting or crossover manoeuvres during various sport disciplines. The way APAs initiate and organise the movement (manoeuvrability)

influences the performance (Jindrich & Qiao, 2009). For instance, it has been reported that higher skilled rugby players demonstrate faster changes of direction than less skilled players (Gabbett & Benton, 2009).

But manoeuvrability is also associated with injury risk (Jindrich & Qiao, 2009). Indeed, lateral movements with changes of direction may lead to misalignments in the lower extremities with the consequence that the knee may be at a relatively high risk of injuries, such as anterior cruciate ligament ruptures (Alentorn-Geli et al., 2009; Olsen, Myklebust, Engebretsen, & Bahr, 2004). As football is the most commonly practiced sport in the world, the injury concern in this activity is even higher than in other sports (Alentorn-Geli et al., 2009), which underlines the need for researchers to keep improving the body of knowledge dealing with knee injury mechanisms and risk factors associated with football practice. Moreover, knee injuries are often experienced in an unpredictable setting, as for instance perturbed landing or insufficient time to prepare the movement (Olsen et al., 2004). Experimental results revealed that knee joint loading, especially knee valgus moment, was greater during unpredictable conditions and consequently may have increased the risk of knee injuries during these cutting manoeuvres (Besier, Lloyd, Ackland, & Cochrane, 2001; Houck, Duncan, & Haven, 2006).

Therefore, the influence of unpredictable conditions on APAs during cutting manoeuvres and their consequences for knee injury risk are of interest. Increasing the body of knowledge dealing with APAs would help to further understand how performance and safety could be improved during such tasks typically encountered during football games. In addition to the unpredictability of the upcoming change of direction, the time available to prepare the movement plays an important role as recently proposed by Brown, Palmieri-Smith and McLean (2009). Although APA strategies have been fairly well described during changes of direction, their limitations have not been investigated so far during critical situations, i.e. during a cutting manoeuvre with little time available to prepare.

The aim of the present study was to determine which APAs are the most relevant during cutting manoeuvres in football and which fail when time available to prepare the movement is limited. We hypothesised that football players would increase their lateral trunk flexion when the time available to prepare the cutting manoeuvre was reduced. Finally, we aimed to evaluate the consequences for knee injury risk resulting from these APA failures.

Materials and methods

Participants

Using data from the literature evaluating unanticipated movements (Brown et al., 2009), the sample size was estimated to achieve 80% statistical power with an alpha level of 0.05. Accordingly, 13 male amateur football players (age: 24 ± 3 years old; height: 1.79 ± 0.03 m; mass: 73.8 ± 7.4 kg; football experience: >10 years in the 5–7th German league) participated in the study. Prior to testing, all participants were informed about possible risks and gave written informed consent. Participants did not have a previous history of serious knee injury or any current knee pain. All procedures were approved by the local ethics committee.

Protocol

Football players were asked to perform three different tasks in a randomised order, including a cutting manoeuvre to 45° , a crossover to -20° and a straight run with an approach running speed of 5 ± 0.2 m \cdot s $^{-1}$. The tasks were performed on a force plate and were followed by a maximal sprint over a 4 m distance.

The direction of the movement was indicated by a light signal occurring at different times before the left foot contacted the force plate. The first condition was anticipated (Ant), meaning that participants already knew which light would turn on before starting to run. For this condition the light stimulus appeared 850 ms before ground contact. The three other cutting conditions were performed without this knowledge, while the light stimulus was presented 850, 600 and 500 ms before ground contact. The rationale for these delays was determined during pilot measurements. 850 ms was considered long enough to prepare the movement in a similar manner to the Ant condition. In this condition, the light stimulus appeared approximately three steps (during right foot contact) before the change of direction. 600 ms was felt by the participants to be a typical time constraint comparable to game situations and the signal occurred during the initial contact of the left foot two steps prior to changing the direction. From this value, we decreased the time available until players started to fail in the task execution, but not in more than 20% of the trials. In this condition (i.e. 500 ms condition), the light stimulus occurred during the push off phase of the left foot contact two steps prior to the change of direction.

The order of the four time conditions was randomly assigned over the 13 participants. Participants performed the movements on an artificial football turf (Ecofill®, Mondo S.p.A., Alba,

Italy) while wearing the same neutral football shoe (Predator, Adidas, Herzogenaurach, Germany). All participants were used to such movements due to their football practice and were also familiar with the light stimulus procedure, as they had already participated in studies using this testing set-up.

During the present study, participants had to perform about 80 running and cutting movements to ensure eight successful 45° cutting manoeuvres for each time condition. Participants had at least 30 s recovery after each trial but could rest as long as they wanted to avoid the onset of fatigue.

The 45° cutting manoeuvre was the only movement to be further analysed due to its relevance when investigating performance and injury aspects of changing direction tasks (see Introduction). A trial was defined as successful if the approach speed was reached, the left foot hit the force plate and the sprint was conducted in the direction of the light stimulus already one step after the force plate.

Material

The approach running speed was measured over the last 3 m before the force plate (Timer S3, Alge-Timing, Palling, Germany) and the sprint velocity over the first 3 m after the force plate were recorded by means of light switches (M18 series, Panasonic Electric Works Europe AG, Holzkirchen, Germany).

The visual light stimulus was triggered automatically via a light switch (M18 series, Panasonic Electric Works Europe AG, Holzkirchen, Germany), triggered during the approach run. The three lights were placed at eye level in the directions of the three tasks, 7 m away from the force plate.

Three-dimensional head, trunk and leg kinematics were recorded using reflective skin markers (Ø 14 mm) placed with self-adhesive tape on anatomical landmarks of the participants' head, trunk, pelvis and both right and left legs. Specifically, markers were placed around the head (four markers), on the suprasternal notch and the transition between the body of sternum and the xiphoid process as well as on the T6 vertebra, on the anterior and posterior superior iliac spines, on the great trochanter, medial and lateral epicondyle of the knee, tibia (two markers) and the medial and lateral malleolus of the left leg, and on the posterior, medial and lateral aspects of the heel cup as well as the toe of both feet. This marker placement was based on previously published articles for the head (Sreenivasa et al., 2008) and the lower limb (Fleischmann, Gehring, Mornieux, & Gollhofer, 2010). The marker positions for the trunk segment were developed based on the conclusions reported by Leardini, Biagi, Belvedere and Benedetti (2009). Markers were captured with a 12-camera motion analysis

system (Vicon V-MX, VICON Motion Systems Ltd., Oxford, UK) at a sampling frequency of 200 Hz. A static trial with the participant standing in a pre-defined neutral position was used to define the kinematic model and to calculate segment lengths and joint centres (Grood & Suntay, 1983; Wu et al., 2002). Joint kinematics in three rotational degrees of freedom were determined using an yxz Euler rotation sequence of the respective segment coordinate system. Marker trajectory and ground reaction force (BP600900, AMTI, Watertown, USA) signals were both filtered with a low pass Butterworth filter (4th order, 20 Hz cut-off frequency) prior to calculating external joint moments with a standard inverse dynamics approach.

Data analysis

Kinematic parameters were first investigated at the end of the right foot contact, i.e. one step prior to the change of direction, as the body posture at this moment would represent the APAs and would also help to understand the organisation of the transition phase between APAs and the moment of the change of direction. The right heel strike and toe off were determined based on the peak vertical acceleration signal of the heel marker and toe marker, respectively. Head rotation, trunk rotation and lateral trunk flexion with respect to the global coordinate system as well as step width were calculated (for positive orientation see Figure 1).

Then the kinematic parameters were analysed at the time of the peak knee abduction moment during the left foot contact during the cutting manoeuvre, as this is a relevant parameter associated with knee injury (Hewett et al., 2005). Again, head rotation, trunk rotation and lateral trunk flexion as well as step width were determined. In addition, the foot progression angle was defined as the rotation of the calcaneus in relation to the global coordinate system. Moreover, hip and knee joint moments in the frontal plane were analysed.

Statistics

The selected biomechanical parameters were averaged across eight trials for each participant in each cutting condition and served as the basis for the statistical analysis (SPSS 20.0, IBM SPSS Statistics, IBM Corporation, Armonk, NY, USA). After having confirmed that the data followed a normal distribution (Kolmogorov–Smirnov test), main effects of cutting condition (Ant vs. 850 ms vs. 600 ms vs. 500 ms) on the dependent biomechanical variables were tested using a one-way analysis of variance (General Linear Model) with repeated measures. Pearson's correlation coefficient was used to

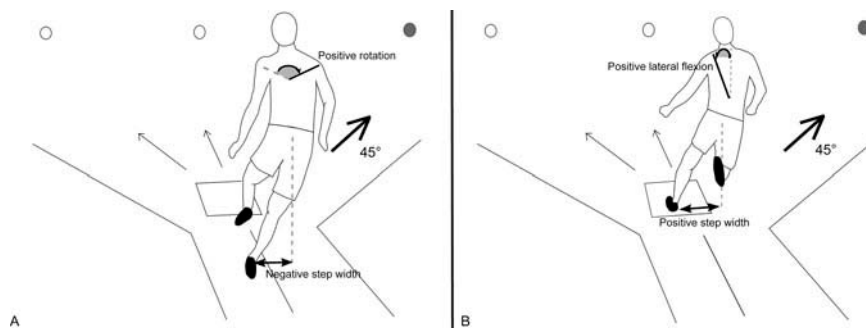


Figure 1. Schematic representation of one participant performing the 45° cutting manoeuvre to the right in response to the grey light stimulus. Figure A represents the end of the right foot contact, one step prior to the change of direction and Figure B the time of the peak knee abduction moment during the left foot contact.

Notes: Positive rotation values of the head, trunk and foot progression angle are orientated towards the cutting direction. Positive lateral trunk flexion values indicate a tilt of the trunk to the left. Positive step width values define a lateral foot placement with respect to the vertical projection of the hip joint centre.

determine the relation between lateral trunk flexion and peak knee abduction moment during the cutting manoeuvre (Jamison, Pan, & Chaudhari, 2012). The level of significance was set at 0.05. If a statistically significant main effect was observed, least significant difference (LSD) multiple comparison procedures determined where the differences occurred. All results are presented as group means \pm s.

Results

One step before changing the direction, there was a significant influence of the cutting condition on head rotation ($P = 0.033$), trunk rotation ($P = 0.002$) and lateral trunk flexion ($P < 0.001$), whereas step width was not reduced ($P = 0.081$) (see Table I).

In the cutting condition with 600 ms the head was significantly less rotated to the right compared to both Ant and 850 ms ($P = 0.033$ and $P = 0.028$, respectively), while higher trunk rotation to the left was found for both 600 and 500 ms compared to Ant and 850 ms ($P < 0.026$) and higher lateral trunk flexion values were reported for both 600 and 500 ms than Ant and 850 ms ($P < 0.001$). None of these parameters were significantly different between Ant and 850 ms, as well as between 600 and 500 ms.

At the time of the peak knee abduction moment, there was a significant influence of the cutting condition on head rotation ($P = 0.015$), trunk rotation ($P = 0.028$) and lateral trunk flexion ($P < 0.001$), whereas step width and foot progression angle were not altered (Table II).

During the cutting condition with 600 ms the head was significantly less orientated towards the right compared to both Ant and 850 ms ($P = 0.032$ and $P = 0.018$, respectively). The 500 and 600 ms cutting conditions yielded larger trunk rotation to the left than 850 ms ($P = 0.036$ and $P = 0.043$, respectively). Compared to both Ant and 850 ms, lateral trunk flexion values were significantly higher for 600 ms ($p < 0.002$) and 500 ms ($p < 0.001$). Additionally, the lateral trunk flexion was higher for 500 than 600 ms ($P = 0.015$). None of these parameters were significantly different between Ant and 850 ms.

Mean values of the parameters dealing with the loading of the lower limb at the time of the peak knee abduction moment are reported in Table III. Along with the reduction of the time to prepare the cutting manoeuvre, there was a significant increase of the posterior ($P < 0.001$) and vertical ground reaction forces ($P < 0.001$). The cutting condition significantly influenced both hip ($P < 0.001$) and knee

Table I. Mean \pm s values of the anticipatory postural strategies one step prior to the change of direction for the four cutting manoeuvre conditions (Ant, 850 ms, 600 ms, 500 ms). Positive rotation values are orientated to the cutting direction, positive lateral flexion values are towards the left and negative step width values define a medial foot placement.

	Ant	850 ms	600 ms	500 ms
Head rotation (°)	11.1 \pm 9.1	10.9 \pm 9.9	6.5 \pm 10.5 ^{*a}	6.9 \pm 7.1
Trunk rotation (°)	-7.1 \pm 8.4	-7.0 \pm 8.5	-10.8 \pm 9.3 ^{*a}	-11.1 \pm 7.3 ^{*a}
Lateral trunk flexion (°)	1.9 \pm 5.2	3.6 \pm 4.7	9.8 \pm 5.4 ^{*a}	12.2 \pm 5.7 ^{*a}
Step width (m)	-0.24 \pm 0.06	-0.27 \pm 0.08	-0.23 \pm 0.09	-0.21 \pm 0.11

Note: ^{*a} significant differences compared to Ant and 850 ms cutting conditions ($P < 0.05$).

Table II. Mean \pm s values of the postural strategies at the time of the peak knee abduction moment for the four cutting manoeuvre conditions (Ant, 850 ms, 600 ms, 500 ms).

	Ant	850 ms	600 ms	500 ms
Head rotation (°)	18.0 \pm 11.3	17.6 \pm 13.1	13.3 \pm 13.5 ^{*a}	13.5 \pm 11.5
Trunk rotation (°)	-4.8 \pm 11.0	-3.3 \pm 11.1	-7.3 \pm 10.3 ^{*b}	-8.8 \pm 9.9 ^{*b}
Lateral trunk flexion (°)	5.3 \pm 6.1	6.1 \pm 6.4	10.4 \pm 6.4 ^{*a}	13.7 \pm 6.9 ^{*a,c}
Step width (m)	0.36 \pm 0.03	0.35 \pm 0.04	0.36 \pm 0.03	0.35 \pm 0.03
Foot progression angle (°)	7.8 \pm 7.8	9.2 \pm 7.9	6.9 \pm 9.7	6.3 \pm 9.8

Notes: Positive rotation values are orientated to the cutting direction, positive lateral flexion values are towards the left and positive step width values define a lateral foot placement.

* expresses a significant difference.

^a compared to Ant and 850 ms cutting conditions.

^b compared to the 850 ms cutting condition.

^c compared to the 600 ms cutting condition.

Table III. Mean \pm s values of the ground reaction forces (posterior and vertical), hip and knee abduction moments at the time of the peak knee abduction moment and of the contact time for the four cutting manoeuvre conditions (Ant, 850 ms, 600 ms, 500 ms).

	Ant	850 ms	600 ms	500 ms
Posterior force (N \cdot kg ⁻¹)	5.42 \pm 2.51	6.11 \pm 2.56	7.38 \pm 2.25 ^{*a}	8.03 \pm 2.16 ^{*a,c}
Vertical force (N \cdot kg ⁻¹)	15.8 \pm 5.9	17.1 \pm 7.0	19.7 \pm 5.7 ^{*a}	20.8 \pm 5.9 ^{*a}
Hip abduction moment (N \cdot m \cdot kg ⁻¹)	2.01 \pm 0.78	2.20 \pm 0.88	2.55 \pm 0.74 ^{*a}	2.57 \pm 0.81 ^{*a}
Knee abduction moment (N \cdot m \cdot kg ⁻¹)	1.00 \pm 0.48	1.04 \pm 0.48	1.21 \pm 0.47 ^{*a}	1.24 \pm 0.49 ^{*a}
Contact time (s)	0.22 \pm 0.03	0.22 \pm 0.03	0.24 \pm 0.03 ^{*a}	0.25 \pm 0.04 ^{*a,c}

Notes: * expresses a significant difference.

^a compared to Ant and 850 ms cutting conditions.

^c compared to the 600 ms cutting conditions

abduction moments ($P < 0.001$). Moreover, the contact time was influenced by the cutting condition ($P < 0.001$).

Again, none of these parameters were significantly different between Ant and 850 ms. Compared to Ant and 850 ms, posterior ground reaction force was significantly higher for 600 ms ($P = 0.005$ and $P = 0.024$, respectively) and 500 ms ($P < 0.001$ for both), while 500 ms even yielded a higher posterior ground reaction force than 600 ms ($P = 0.04$). The vertical ground reaction force during the cutting condition with 600 ms was significantly higher than Ant and 850 ms ($P = 0.003$ and $P = 0.015$, respectively). In the same manner, performing the cutting manoeuvre with 500 ms resulted in higher vertical ground reaction forces than both Ant ($P = 0.001$) and 850 ms ($P = 0.006$). The 500 ms cutting condition yielded increased hip abduction moment compared to Ant ($P = 0.002$) and 850 ms ($P = 0.007$), as well as increased knee abduction moment compared to Ant ($P = 0.001$) and 850 ms ($P = 0.003$). The same observation could be made for 600 ms, which, compared to Ant and 850 ms, generated higher hip and knee abduction moments ($P < 0.009$ and $P < 0.011$, respectively). Compared to Ant and 850 ms, contact time values were significantly higher for 600 ms ($P = 0.007$ and $P = 0.002$,

respectively) and 500 ms ($P < 0.001$ for both). Additionally, the contact time was longer for 500 than 600 ms ($P = 0.008$).

To further understand the influence of lateral trunk flexion on knee abduction moment, variations from the Ant condition baseline in the three unpredictable cutting conditions for these two parameters were plotted together (Figure 2). Lateral trunk flexion significantly correlated with peak knee abduction moment ($r = 0.41$; $P = 0.009$).

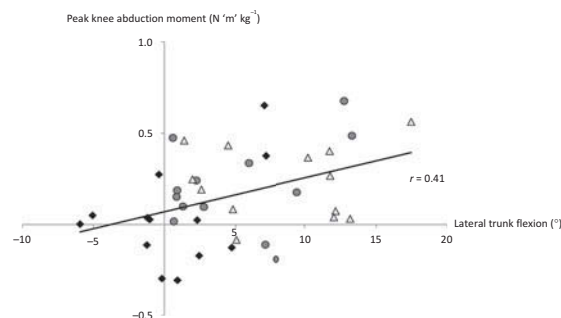


Figure 2. Linear regression between peak knee abduction moment and lateral trunk flexion during the 45° cutting manoeuvres. Values represent the difference from the Ant condition for the 850 ms (black diamonds), 600 ms (grey circles) and 500 ms (grey triangles) cutting conditions.

Sprint velocity remained comparable over the four cutting conditions, ranging from $5.0 \pm 0.4 \text{ m} \cdot \text{s}^{-1}$ for Ant to $4.9 \pm 0.3 \text{ m} \cdot \text{s}^{-1}$ with a delay of 500 ms.

Discussion

The primary finding of the present study was that APAs are modulated according to the time available to prepare a cutting manoeuvre. Participants relied on a higher lateral trunk flexion to perform cutting manoeuvres under limited time available. While this was an effective strategy to maintain the sprint performance, higher lateral trunk flexion was associated with increased knee abduction moment. Finally, as soon as the time available was sufficient (850 ms), not knowing in advance in which direction one would have to cut did not influence APAs, knee joint loading or sprint performance.

Of the three main APA strategies proposed in the literature, lateral trunk flexion was the most influenced by the cutting condition. Head and trunk rotations were also influenced by the time available to prepare the cutting manoeuvre, whereas the step width regulation was not significantly altered.

With less time available, lateral trunk flexion increased by 10.3° (500 ms vs. Ant) already one step prior to the cutting manoeuvre and by 8.4° at the time of the change of direction 150 ms later. This finding is in agreement with data from previous studies (Houck et al., 2006; Patla et al., 1999). Lateral trunk flexion was even more increased in the present study compared to the $+3.7^\circ$ found by Houck et al. (2006) when participants performed a 50° cutting task under unpredictable conditions but with a lower approach velocity of $1.9 \text{ m} \cdot \text{s}^{-1}$. The higher lateral trunk flexion can be seen as a strategy to move the CoM in the new direction. A neuromuscular explanation for this increased lateral trunk flexion strategy could be the relative higher strength of the trunk and hip muscles compared to other rather weak muscles (e.g. ankle invertors/evertors) during the stance phase. Therefore, participants could rely more on their trunk than on any foot placement strategy.

Head and trunk rotations were influenced to some degree by the time restriction to prepare the cutting manoeuvre. With less time available, the head was less orientated towards the cutting direction and the trunk was even more rotated in the opposite direction at the time of changing the direction as well as one step before. The rotation of the trunk in the opposite direction has already been reported in the literature during cutting manoeuvres (Dempsey et al., 2007; Dempsey, Lloyd, Elliott, Steele, & Munro, 2009). The influence of the cutting condition on head and trunk rotations was limited, as for instance no significant difference was observed

between the 500 ms and Ant conditions at the time of the peak knee abduction moment or given that standard deviation values for the head ($7\text{--}13^\circ$) and the trunk ($7\text{--}11^\circ$) were large. The relatively small influence of the time available on these APAs further shows the weak importance of head and trunk rotation strategies during cutting manoeuvres.

The step width strategy depends on the foot placement. One step prior to changing the direction, a medial shift of the foot placement with respect to the CoM would allow the CoM to accelerate towards the new direction. However, this strategy tended to be less used under critical time conditions ($<600 \text{ ms}$). Moreover, the subsequent foot contact when changing the direction was not modulated, i.e. the foot was not placed more laterally to increase the acceleration of the CoM in the cutting direction. This result is in agreement with Patla et al. (1999) and partially consistent with Wheeler and Sayers (2010) and Houck et al. (2006). Indeed, as soon as the change of direction was unpredictable, the first authors reported that the step width strategy was significantly less used one step prior to changing the direction and the latter authors found a reduction of the step width at the time of the cutting manoeuvre. Moreover, less use of the step width strategy one step before the change of direction was associated with reduced lateral movement speed (Wheeler & Sayers, 2010). Therefore, the step width strategy does not seem to be the one to increase or to rely on during challenging cutting manoeuvres.

Accordingly, altering lateral trunk flexion is probably the most relevant postural strategy to successfully perform a cutting manoeuvre under unpredictable conditions and to maintain the subsequent sprint performance at the same level.

Despite obvious main effects of the cutting condition on APAs, it is worth noting that further statistical analysis revealed no significant difference between Ant and 850 ms. Accordingly, not knowing in advance which cutting manoeuvre will have to be executed does not influence APA strategies as long as the time available to prepare the movement is long enough. Under the present conditions, i.e. $5 \text{ m} \cdot \text{s}^{-1}$ approach velocity and 45° cut angle, a time of 850 ms (three steps prior to the change of direction) was enough to use postural strategies comparable to those observed under predictable conditions.

Whereas lateral trunk flexion was increased during successful cutting manoeuvres under limited time to prepare the movement, the overall leg loading as well as the joint loading increased. Indeed, posterior and vertical ground reaction forces at the time of the peak knee abduction moment (roughly 20–25 ms after foot strike) were significantly higher for both the 600 ms and the 500 ms cutting conditions. This

higher loading of the leg could increase the difficulty of the task, requiring significantly longer contact time and explaining also why 20% of the cutting manoeuvres with 500 ms time available were not successful.

Moreover, hip and knee joint moments increased in the frontal plane as soon as the time available decreased to 600 ms. This observation is in agreement with previous results showing increased joint loading under unpredictable conditions (Besier et al., 2001; Brown et al., 2009), although Dempsey et al. (2009) reported no significant influence of unpredictability on knee joint moments under conditions comparable to the present study (approach velocity, cutting angle and light stimulus timing). The large peak knee abduction moment values (up to $1.24 \text{ N} \cdot \text{m} \cdot \text{kg}^{-1}$) are consistent with recently published data (Vanrenterghem, Venables, Pataky, & Robinson, 2012) and underline the high loading of the knee and the possible risk for knee injury. The substantial new information provided by the present study is that not only cutting angle or approach velocity or even the unpredictability about the cutting condition influence knee joint loading but also the time available to prepare the cutting manoeuvre.

The increased lateral trunk flexion due to the reduction of the time available explained approximately 17% of the variance in peak knee abduction moment and should therefore be considered as an important parameter when investigating knee injury mechanisms (Figure 2). This association confirms previous observations dealing with the mechanical coupling between trunk and knee joint in the frontal plane (Dempsey et al., 2009; Hewett & Myer, 2011; Houck et al., 2006; Jamison et al., 2012).

Thus, it can be concluded that altered APA strategies under unpredictable conditions are one factor leading to an increased knee joint injury risk during cutting manoeuvres, along with, for instance, less suitable neuromuscular activity of the knee joint muscles (Besier, Lloyd, & Ackland, 2003).

Accordingly, the aim of future studies could be to provide solutions for this excessive use of lateral trunk flexion. Reducing the use of this strategy would then probably need to be compensated by other anticipatory postural strategies to maintain the subsequent sprint performance at the same level. For instance it is known that head and trunk rotation strategies tend to be reduced with maturation (Vallis & McFadyen, 2005). Therefore, training these strategies again to improve the re-orientation of the whole body based on a head and trunk rotation might be a relevant solution. Another solution could be a modified foot placement. Dempsey et al. (2009) have already reported a reduction of knee joint loading associated with a specific whole body cutting

technique, characterised by a medial shift of the foot together with a more neutral lateral trunk flexion at the time when changing the direction. However, even if a more externally rotated foot placement, i.e. the foot progression angle, might be a strategy to reduce knee joint frontal plane loading (Fleischmann et al., 2010; Sigward & Powers, 2007), this parameter was not influenced by the cutting conditions in the present study. Therefore, modulation of the foot rotation does not seem to be an appropriate strategy to prepare a cutting manoeuvre under the limited time available.

It has to be emphasised that any cutting technique modification through a training intervention would probably be more effective if attention is brought to the step prior to the change of direction, whichever strategy, i.e. head rotation, trunk rotation or step width regulation, is trained.

The relevance of the present results can be transferred to daily football training. First, training should put emphasis on decision making situations, where players have to identify ball trajectories, teammate or opponent displacements that are not pre-defined and make the appropriate decision accordingly (relevant timing and direction of the pass or displacement). Second, training should focus on situations where players experience one against one situations (either by attacking or defending) and where changes of direction are required. During these situations, players would focus on their trunk rotation and the lateral trunk flexion to operate such cutting manoeuvres. The upper body should not only be kept straight but also orientated towards the new direction of the cutting manoeuvre. Moreover, players should pay attention to the alignment of the leg and the upper body already one step prior to the change of direction.

To conclude, this study provides a detailed biomechanical and functional analysis of APA strategies during cutting manoeuvres in football. From the variables analysed in the present study, increasing lateral trunk flexion is the most effective strategy to successfully perform a cutting manoeuvre when less time is available to prepare the movement. But this is also associated with increased knee injury risk. Considering knee injury as a serious issue in football, training programs need to improve APAs to reduce injury risk and possibly enhance performance.

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Medial Compressible Forefoot Sole Elements Reduce Ankle Inversion in Lateral SSC Jumps

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Sideward movements are associated with high incidences of lateral ankle sprains. Special shoe constructions might be able to reduce these injuries during lateral movements. The purpose of this study was to investigate whether medial compressible forefoot sole elements can reduce ankle inversion in a reactive lateral movement, and to evaluate those elements' influence on neuromuscular and mechanical adjustments in lower extremities. Foot placement and frontal plane ankle joint kinematics and kinetics were analyzed by 3-dimensional motion analysis. Electromyographic data of triceps surae, peroneus longus, and tibialis anterior were collected. This modified shoe reduced ankle inversion in comparison with a shoe with a standard sole construction. No differences in ankle inversion moments were found. With the modified shoe, foot placement occurred more internally rotated, and muscle activity of the lateral shank muscles was reduced. Hence, lateral ankle joint stability during reactive sideward movements can be improved by these compressible elements, and therefore lower lateral shank muscle activity is required. As those elements limit inversion, the strategy to control inversion angles via a high external foot rotation does not need to be used.

Keywords: shoes, ankle sprains, foot progression angle, kinematics, kinetics, EMG

Lateral movements are central components in such sports as basketball, handball, soccer, volleyball, and tennis. They occur in complex motions, such as in cuttings or in side shuffles. These lateral movements have in common that they are generally realized as a reactive stretch-shortening cycle (SSC) conducted to the side. The SSC describes a combined muscle action in which the preactivated muscle is stretched during landing, immediately followed by a concentric push-off contraction.^{1,2} As the lower extremity joints' anatomy and the joints' encompassing muscles provide less potential for stabilization in the frontal plane,³ these lateral SSCs are biomechanically rather difficult to control compared with sagittal plane movements.

Thus, excessive frontal plane ankle joint inversions during lateral SSC movements often result in lateral ankle sprains. Ankle sprains occur when the external inversion moments acting on the ankle joint (1) are greater than the counteracting internal eversion moments provided by the ankle joint's encompassing muscles and ligaments, and (2) when these external inversion moments consequently cause a further rapid acceleration of inversion beyond the normal range of motion.⁴ Those injuries represent the most common peripheral joint injury.^{5,6} A

movement analysis of ball games showed that basketball had the highest frequency of movements related to the occurrence of ankle sprains: lateral motions, rotations, jumping, and landings.⁷ Long-term analysis of American collegiate basketball also shows that approximately 1/4 of all injuries in this sport are ankle sprains.^{8,9}

Thus, in particular in basketball, but also in other court sports with high incidences of ankle sprains, external devices such as braces and tapes are applied to the ankle to laterally stabilize the joint and to reduce ankle joint injuries.^{10,11} Besides these measures, special sport shoe construction characteristics try to improve the lateral ankle joint stability. The lateral stability of a sport shoe can be defined as the ability to resist inversion.¹² One central measurement to improve the lateral ankle joint stability in basketball is to heighten the shoe shaft to limit the ankle inversion.¹³ But so far only a few studies evaluated basketball shoe sole constructions with respect to ankle joint stability. When the influence of the shoe sole on lateral ankle joint stability is being analyzed, it needs to be considered that soles provide a greater leverage to the frontal plane ankle joint moment compared with a barefooted movement conduction.^{14,15} Therefore, special rearfoot sole constructions were developed to enhance lateral ankle joint stability during frontal plane movements. These constructions should reduce inversion angles and the corresponding moment leverage. Stacoff et al⁴ could, for instance, demonstrate that a rearfoot sole construction which moved laterally under loading increased the foot's inclination and consequently reduced ankle inversion.

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Also a medially banked rearfoot sole allowed a greater foot inclination during sideward landings which suggested a lower ankle inversion in comparison with a shoe with a standard sole design.⁷ But the influence of banked forefoot sole constructions on ankle stability has not been the subject of any investigations.

In addition to external mechanical ankle stabilization measurements, lateral ankle joint stability also depends on neuromuscular control. Thus, in particular the peronei oppose an ankle inversion. The function of these muscles to provide dynamic stability during lateral movements is therefore considered to be important to impede a lateral ankle sprain. Consequently, peroneal reactions to sudden inversion movements were the subject of many studies as reviewed by Menacho et al.¹⁶ In dynamic motor tasks, it has been observed that subjects suffering from chronic ankle instability exhibit reduced peroneus longus activity accompanied by increased ankle inversion during drop jumps.¹⁷ This observation leads to the assumption that peroneus longus activity might be of importance for lateral ankle joint stability during reactive lateral movements. Unfortunately, the presented studies^{4,7} investigating the influence of rearfoot sole modifications on lateral ankle stability did not analyze muscle activity. Hence, so far the effect of sole construction modifications on the muscular ankle joint stability during dynamic lateral movements remains unclear.

A recent study focusing on load adjustment strategies in reactive lateral SSC jumps demonstrated that a greater externally rotated foot placement marks a central strategy to shift loading from the frontal to the sagittal plane, which is less prone to acute lower extremity joint injuries.¹⁸ In gait analysis it could be demonstrated that an external foot rotation compared with a neutral or internally rotated foot positioning reduces ankle inversion as well as inversion moments.^{19,20} As foot placement influences frontal plane ankle joint mechanics, this might be an important parameter in the evaluation of footwear influence on the lower extremity motor control in reactive lateral SSC movements.

As demonstrated by Stacoff et al.,⁷ shoes with rearfoot banking concepts are able to reduce ankle inversion. However, taking into account that dynamic lateral movements are generally characterized by initial medial forefoot contact, forefoot sole constructions and their influence on ankle joint stability in reactive lateral movements should be investigated. Therefore, it was the purpose of the current study to investigate whether medial compressible forefoot sole elements reduce inversion during the landing phase of a reactive lateral movement.

We hypothesized that these sole elements would as a consequence of a lower forefoot inversion reduce ankle inversion in comparison with the reference shoe. We further hypothesized that lower ankle inversion with the shoe containing medial compressible forefoot sole elements would cause reduced lateral shank muscle activity. Our third hypothesis addressed foot placement: lower frontal plane ankle joint loading accomplished by lower ankle inversion would result in a less externally rotated foot placement with the shoe containing the modified forefoot sole. This assumption lead to the final hypothesis: the

different foot placement strategy with this modified shoe will affect mediolateral ground reaction forces.

Methods

Twelve male basketball players (192 ± 4 cm, 87 ± 9 kg, 24 ± 3 y) who were familiar with conducting lateral movements participated in the study. They gave their written consent to participate in the approved study design, which was conducted according to the Declaration of Helsinki. All participants were free of pain on the test day and had no history of severe orthopedic or neurological disorders. They had not sustained an ankle sprain during the previous year.

Subjects performed reactive lateral SSC jumps onto a force plate. From a starting position, subjects jumped laterally with their left leg onto the force plate and back to the starting position. Landing on the force plate was performed one-legged, forefoot first, and orientated on a mark fixed perpendicular to the direction of motion onto the force plate. Subjects were instructed to jump back from the force plate to the starting position as fast as possible, to omit trunk rotations, and to keep their jumping technique consistent throughout the measurements. In pilot measurements, lateral SSC jumps proved to be a suitable approach to functionally evaluate medial forefoot sole designs on lower extremities mechanical and neuromuscular adjustments in dynamic frontal plane movements.

Two different basketball shoe sole properties (size UK 12.5) were tested in the study: (1) the reference shoe (Ref), which represented a common basketball shoe with a standard court shoe sole construction (Figure 1a). (2) The second basketball shoe (Figure 1b) was a prototype with a modified medial forefoot sole construction (modS). Besides this special medial forefoot sole element, both shoes were identical in shape, materials, and fitting. Whereas the sole of the reference shoe was exclusively made of the elastomeric material ethylene vinyl acetate, a special banking element was implemented in the medial forefoot sole of the modified shoe. This banking element consisted of soft synthetic polymer sheathing which formed in the midsole four hollow inner core chambers (Figure 1b). In contrast to the solid and therefore stiffer lateral forefoot and rearfoot sole, these hollow inner core chambers can be easily compressed under pressure. Therefore the forefoot sole of the modified shoe (478 N/mm) had a lower overall forefoot stiffness than the reference shoe (505 N/mm). Under lateral loading these medial forefoot sole elements of the modified shoe are being compressed, and hence foot inclination increases in these conditions. This banking effect is expected to reduce forefoot and consequently ankle inversion during a lateral movement.

To evaluate whether with the modified shoe frontal plane ankle joint mechanics are adjusted differently to stretch-load variation than with the reference shoe, study participants performed lateral SSC jumps with both shoes from four distances. According to Fleischmann et al.,¹⁸ the four jumping distances were determined with respect

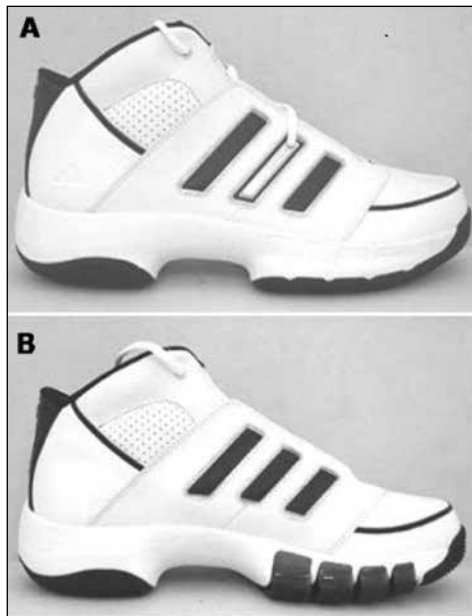


Figure 1 — (A) reference shoe. (B) modified shoe with medial compressible forefoot sole elements.

to the maximum individual lateral jumping distances: Extreme distance (EXD) was set at 100%, long distance (LOD) at 85%, medium distance (MED) at 70%, and short distance (SHD) at 55% of the maximum individual lateral jumping ability.

From each of the four distances, ten lateral jumps were performed. The order of the four jumping distances as well as the order of the shoes were randomized. To exclude learning effects for each footwear condition and jumping distance, the participants could try conducting the movement until feeling comfortable before measurements started. All trials were visually controlled and repeated in the case of an incorrect movement conduction.

Ground reaction forces were measured using a force plate (OR-6-7-2000; Advanced Mechanical Technology Inc, Watertown, MA) with a sampling frequency of 1,000 Hz. Force plate (threshold 10 N) data served as the basis for calculating ankle joint moments and to determine ground contact times. A 3D-motion analysis system (Vicon V-MX, Vicon Motion Systems Ltd., Oxford, UK) consisting of 12 near-infrared cameras with a sampling frequency of 200 Hz was installed to record the kinematics of the subjects' left leg. Retro-reflective markers (14 mm) were placed on anatomical landmarks of the pelvis and the lower left extremity—more precisely, onto the anterior superior iliac spines, the posterior iliac spines, the great trochanter, the medial and lateral epicondyles, two markers onto the tibia, and on the medial and lateral malleolus. Additional markers were externally fixed onto identical positions on the left shoe of the two tested shoe models. Three of these shoe markers were fixed onto the medial, lateral, and upper forefoot cap, and another three markers onto the medial, lateral, and posterior aspect of the lower heel cap.

Kinematic and kinetic data were analyzed using Vicon motion analysis software (Nexus 1.3, Vicon Motion Systems Ltd., Oxford, UK). Calculations were performed according to recommendations of Grood and Suntay²¹ and Wu et al.²² Joint moments were computed using a standard inverse dynamic approach.

For the purpose of this study, the first 25% of ground contact characterizing the early landing (early deceleration) phase were considered. Special emphasis was put to this phase because ankle sprains typically occur during deceleration shortly after touch-down.²³ Thus, mean vertical and mediolateral ground reaction forces of the first 25% of ground contact were calculated. To investigate the influence of the shoe sole characteristics on lateral ankle joint stability, frontal plane joint mechanics were analyzed. Maximum inversion which takes place during that period of ground contact was determined. In addition, the inversion angle at touch-down and the maximum ankle inversion moment were analyzed.

Foot placement was determined by the foot progression angle. The foot progression angle describes the foot's transversal orientation in relation to the global coordinate system. Hence, this angle characterizes the deflection of the foot's longitudinal axis from the landing position perpendicular to the direction of motion. A negative foot progression angle has been defined as internal rotation and a positive angle as external rotation. The foot progression angle was analyzed at the moment of touchdown, to investigate possible footwear-dependent differences in the foot placement strategy initiated before ground contact. The foot progression angle at 25% of ground contact was determined to analyze foot placement after the heel was lowered to the ground.

Bipolar surface electrodes (Blue Sensor, Ambu, Balerup, Denmark) were used to record EMG activity of the shank muscles. Muscle activity of the following left shank muscles was analyzed: gastrocnemius medialis, gastrocnemius lateralis, soleus, peroneus longus, and tibialis anterior. The reference electrode was placed on the medial anterior aspect of the tibia. Skin preparation and electrode placement followed SENIAM recommendations.²⁴ EMG signals were sampled at 1,000 Hz.

EMG data were processed by a LabVIEW-based (National Instruments, Austin, USA) software tool (IMAGO, University of Freiburg, Freiburg, Germany). Data were bandpass filtered (Butterworth, 2nd order, 10–500 Hz), full-wave rectified and integrated (IEMG). With respect to ankle joint stability during the landing phase, the IEMG of the shank muscles was calculated between touchdown and 60 ms of ground contact. This interval was in previous studies used to evaluate ankle inversion in lateral movements.⁴

A two-factor analysis of variance for repeated measurements was calculated to analyze possible dependent variables' differences between the two shoes under the four stretch-load conditions. An alpha level of 0.05 was selected to identify statistical significance.

If a main effect on a dependent variable between the two movements could be observed, differences between

the two shoes concerning this parameter were further analyzed by paired-samples *t* tests for each stretch-load condition. Data normal distribution was tested by the Kolmogorov-Smirnov test. Based on the low number of degrees of freedom and the interindividual dependent variable variances, the reported alpha level was used for these tests to exploratively describe the influence of the four stretch-load conditions on the two observed movements.

Statistical calculations were performed with SPSS 15.0 (SPSS Inc., Chicago, USA). Results are presented as group mean values ± standard deviation. For better readability, the results of the paired samples *t* tests, but not the main effect analyzed with the two-factor analysis of variance are incorporated in the graphics. The latter values are presented in the text.

Results

Mean mediolateral ground reaction forces were higher (*P* < 0.001) with the modified shoe than with the reference shoe (Table 1). As Table 1 also indicates, mean vertical ground reaction forces during the first 25% of ground contact did not show differences between the two shoe conditions. Contact times did not differ between the modified shoe with the medial compressible forefoot sole elements and the reference shoe.

The shoe comparison revealed significantly reduced initial inversion (*P* < .001) and maximum inversion (*P* = .002) with the modified shoe (Figure 2a,b). Figure 2c illustrates that no significant differences for the initial maximum inversion moments between the two shoes could be found.

Foot placement occurred with the modified shoe with a lower external rotation in comparison with the reference shoe. Thus, with the modified shoe the external rotation of the foot at touchdown was lower (*P* < .001) than with the reference shoe (Figure 3a). Also at 25% of the ground contact phase when the heel had been lowered to the ground, with the modified shoe the external rotation was lower (*P* < .001) than with the reference shoe (Figure

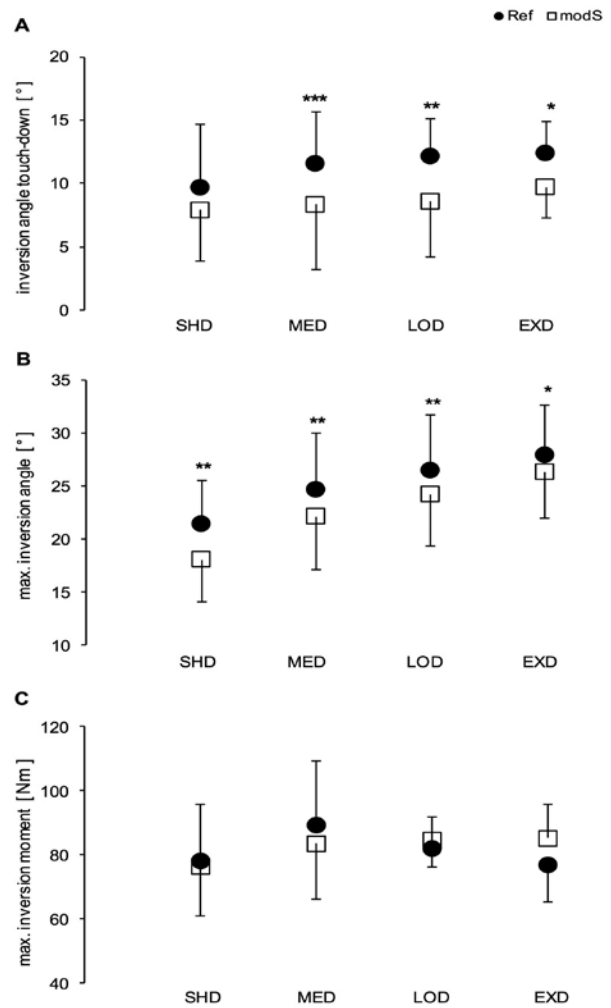


Figure 2 — Inversion angles in dependency of footwear condition and jumping distance. *Ref* = reference shoe, *modS* = modified shoe, *SHD* = short distance, *MED* = medium distance, *LOD* = long distance, *EXD* = extreme distance. (A) angle at touchdown. (B) maximum inversion angle. (C) maximum initial inversion moment **P* < .05; ***P* < .01; ****P* < .001.

Table 1 Ground reaction forces and ground contact times for reference and modified shoes

	SHD	<i>P</i>	LOD	<i>P</i>	MED	<i>P</i>	EXD	<i>P</i>
Mean Mediolateral GRF (N)								
Ref	473.98 ± 90		541.41 ± 117	**	633.84 ± 98	*	703.44 ± 67	**
modS	498.39 ± 100		570.05 ± 101		666.78 ± 102		756.99 ± 87	
Mean Vertical GRF (N)								
Ref	932.11 ± 222		974.78 ± 188		1080.70 ± 218		1138.37 ± 31	
modS	953.92 ± 152		953.06 ± 167		1079.88 ± 202		1179.89 ± 153	
Ground Contact Time (s)								
Ref	0.313 ± .04		0.366 ± .05		0.411 ± .02		0.487 ± .05	
modS	0.323 ± .03		0.369 ± .04		0.423 ± .04		0.493 ± .07	

Note. *Ref* = reference shoe, *modS* = modified shoe, *SHD* = short distance, *MED* = medium distance, *LOD* = long distance, *EXD* = extreme distance, *GRF* = ground reaction force. **P* < .05; ***P* < .01.

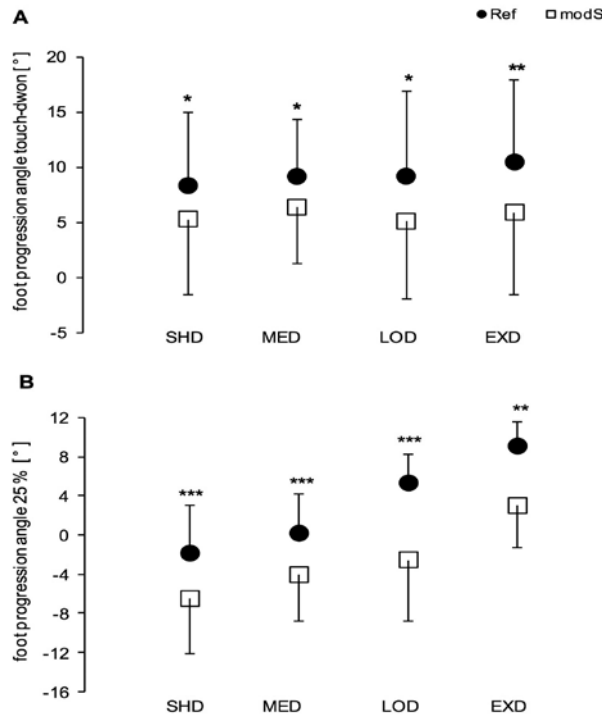


Figure 3 — Foot progression angle in dependency of footwear condition and jumping distance. *Ref* = reference shoe, *modS* = modified shoe, *SHD* = short distance, *MED* = medium distance, *LOD* = long distance, *EXD* = extreme distance. (A) at touchdown. (B) at 25% of ground contact. * $P < .05$; ** $P < .01$; *** $P < .001$.

3b). The foot progression angle during ground contact of a lateral SSC jump is shown in Figure 4 for one subject. The graphic additionally illustrates the parameters inversion angle, inversion moment, and the mediolateral ground reaction force.

The shoe comparison revealed a tendency to lower muscle activities during the initial landing phase with the modified shoe (Figure 5). IEMGs were significantly reduced in the lateral shank muscles peroneus longus ($P < .021$) and gastrocnemius lateralis ($P < .007$). Figures 5b and d indicate that the differences in muscle activation tended to be smaller in the longer jumping distances than in the shorter ones. The remaining shank muscles gastrocnemius medialis, soleus and tibialis anterior did not show a significant activation difference in the shoe comparison.

Discussion

High ankle inversion during the initial landing phase of dynamic frontal plane movements mark a potential risk for sustaining lateral ankle sprains. Therefore, special court shoes are developed which try to reduce that injury risk. The implementation of medial compressible forefoot elements into the sole construction reduced inversion in lateral SSC jumps. In addition, muscle activity of the lateral

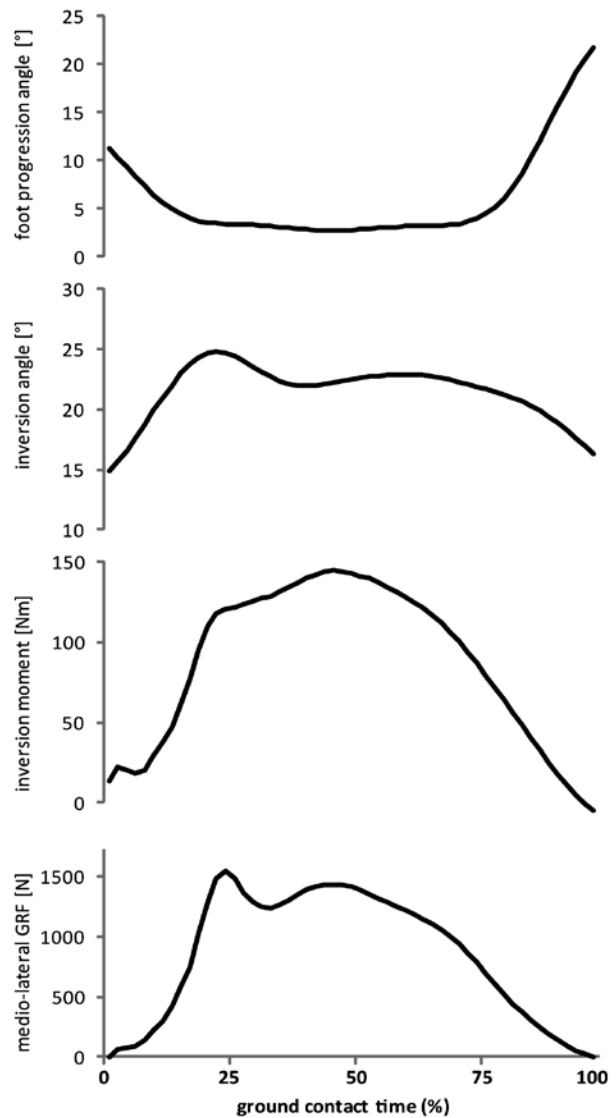


Figure 4 — Single representative trial (medium distance) showing the foot progression angle, the inversion angle, the inversion ankle moment, and the mediolateral ground reaction force over the ground contact.

shank muscles was lower, and foot placement occurred with a reduced external rotation with this modified shoe.

Medial compressible forefoot sole elements reduce the inversion angle at touchdown and the maximum inversion during the landing phase of lateral SSC jumps in comparison with a shoe with a standard sole construction. Incorporating medial compressible forefoot elements into shoe construction might therefore contribute to reducing the risk to sustain an ankle sprain during dynamic lateral movements. The potential to limit inversion in lateral movements by special sole designs has previously been shown for rearfoot sole modifications. In a study analyzing cuttings, it could be demonstrated that soft compressible rearfoot sole constructions minimize inversion in comparison with a harder and less compressible sole.⁴

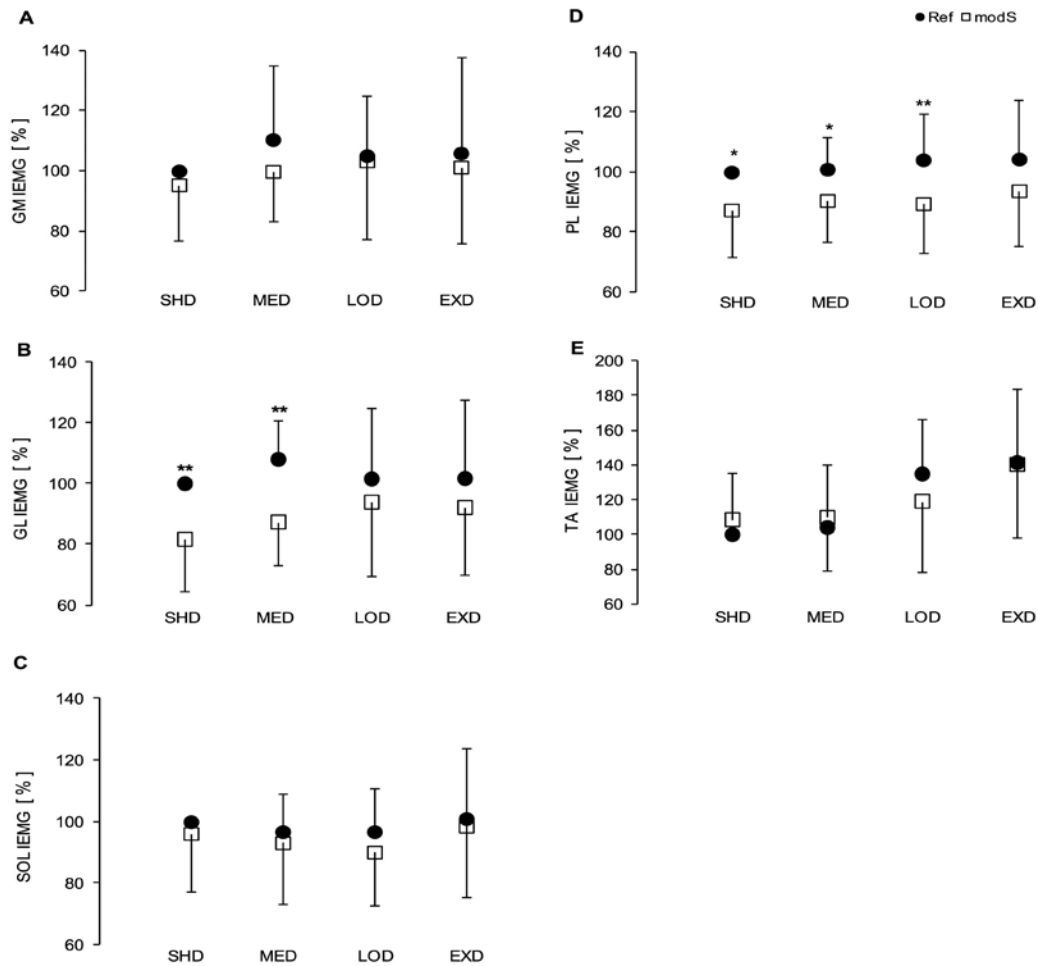


Figure 5 — Lower leg muscles' IEMGs during the initial phase of contact. *Ref* = reference shoe, *modS* = modified shoe, *SHD* = short distance, *MED* = medium distance, *LOD* = long distance, *EXD* = extreme distance. (A) gastrocnemius medialis. (B) gastrocnemius lateralis. (C) soleus. (D) peroneus longus. (E) tibialis anterior. * $P < .05$; ** $P < .01$.

In a functional shoe comparison, it is methodologically necessary to externally mount markers onto the shoes, to be able to analyze foot and ankle kinematics and kinetics. Thus, it needs to be considered that the shoe rather than the foot inversion is being measured. Shoe inversion overestimates the actual foot inversion inside the shoe, as demonstrated by Stacoff et al.⁴ This group determined shoe inversion during sideward cutting maneuvers, as well as the actual heel inversion inside the shoe, by cutting holes into the shoes and placing markers directly onto the skin. In our present study, shoe inversion was measured for both shoes using the same marker set. Therefore, the methodological constraint of measuring the shoe inversion should not limit the shoe comparison analysis. However, we cannot comment on the actual foot inversion inside the tested shoes.

Lower inversion angles with the modified shoe probably cause the lower muscle activity of the lateral shank muscles gastrocnemius lateralis and peroneus longus. Lohrer et al.²⁵ demonstrated that reduced ankle inversion which was achieved by applying tape to the

ankle joint resulted in reduced peroneus longus activity. A tendency of lower muscle activity could in our study also be observed for gastrocnemius medialis and soleus. These differences were rather small, but, according to Komi et al.,²⁶ shoe construction differences are expected to have just small effects on EMG activation in sportive movements. The authors came to this conclusion after studying shoe sole hardness characteristics in running.

Ground contact times indicate that the modified shoe allowed performing lateral SSC jumps with the same reactivity as with the reference shoe, but less muscle activity was needed with the modified shoe to control the landing. The tendency of lower muscle activity with the modified shoe might be beneficial to movement performance as according to Nigg²⁷ an optimal shoe requires reduced muscle work and therewith influences performance positively.

The modified shoe exhibited enhanced mediolateral ground reaction forces which could favor greater inversion moments. An explanation for the higher mediolateral ground reaction forces with the modified shoe can

possibly be found in the more internally rotated foot placement with this shoe. This assumption is backed by a study of Sigward and Powers.²⁸ In an analysis of different cutting techniques, they reported that greater frontal ground reaction forces were associated with a lower external rotation of the foot. Also in a gait analysis investigation by Ho et al,²⁰ it was shown that a greater internally rotated foot placement resulted in higher ground reaction forces, and thus in enhanced frontal plane ankle joint moments. The connection between foot placement and the ankle inversion moment could also be demonstrated by Andrews et al,¹⁹ who showed that increased external foot rotation correlated with reduced ankle inversion moments in gait. Based on these findings and on the fact that with the modified shoe mediolateral ground reaction forces were higher and foot placement occurred with a greater internal rotation, we would have expected higher inversion moments with this shoe. But inversion moments did not differ between the two shoes. We assume that the ability of the medial compressible forefoot sole elements to reduce the inversion angle limited the inversion moment in this shoe. And therefore, despite enhanced mediolateral ground reaction forces with this modified shoe, inversion moments did not differ between the two shoes.

The relation between inversion moments and inversion angles under different foot placement strategies showed also the previously cited investigation by Ho et al.²⁰ This group found that increased inversion moments with an internally rotated foot placement were associated with higher inversion angles in comparison with an externally rotated foot placement. But in contrast to Ho et al's investigation,²⁰ in the current study the medial compressible forefoot sole elements could reduce inversion angle. The assumption that the lower inversion angles achieved by these elements limit the inversion moment is supported by the analysis of the four stretch-load conditions: the modified shoe's ability to reduce inversion angles seems to be limited in extreme condition in comparison with the lower jumping distances. Since mediolateral ground reaction forces were also clearly enhanced with the modified shoe in this extreme load condition, the inversion moment with the modified shoe therefore tended to be higher than with the reference shoe in this extreme condition.

Because medial compressible forefoot sole elements decrease inversion, limiting inversion does not need to be achieved by the strategy of a high external foot rotation. Hence, the safety-orientated externally rotated foot placement can be reduced, and a more internally rotated foot progression angle favoring an effective lateral push-off can be realized with the modified shoe.

To our knowledge only one other study addressed foot placement during a lateral movement in a shoe construction comparison before.⁷ In that study the foot progression angle was analyzed during shod and barefooted side-steps, but no consistent conclusion concerning foot placement could be drawn. Absolute angle values of that study are not comparable with those of the current study, as a different methodological approach was used

to determine the foot progression angle. The authors reported a 1 to 2° externally rotated foot placement with a standard sole construction, whereas foot placement with a medially banked shoe led to a 2 to 3° internally rotated foot position. Although that study could just reveal tendencies concerning the foot progression angle, these results are in line with the findings of the current study. The differences between the investigation conducted by Stacoff et al⁷ and the current study result from varying study designs and from the two different investigated lateral movements. Lateral SSC jumps are characterized by higher dynamics than side-steps; consequently, lateral SSC jumps are more difficult to control. The difficulty to control the lateral landing phase is reflected in a greater externally rotated foot placement. Therefore, the external foot rotation in lateral SSC jumps is higher than in side-steps.

The results of the current study, which are supported by the findings of Stacoff et al,⁷ allow the assumption that shoe sole constructions which limit inversion permit a more internally rotated foot placement during dynamic lateral movements. The current study suggests that with the shoe containing medial compressible forefoot sole elements a more internally rotated foot placement strategy could be anticipated, and that medial compressible forefoot sole elements might also mechanically facilitate a greater internal foot rotation during landing. The higher internally rotated foot placement with the modified shoe already at the instant of touchdown indicates a different anticipated foot placement strategy than with the reference shoe. The assumption that the modified shoe also mechanically favored a more internally rotated foot positioning by allowing a greater foot rotation after touchdown into the subsequent push-off direction is supported by the fact that the differences in foot progression angles between the two shoes are more distinct at 25% of ground contact compared with the moment of touchdown.

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