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# Analyse des effets d'une modification expérimentale de la capacité posturo-cinétique induite par deux types d'exercices physiques sur l'organisation biomécanique de la marche et de l'initiation de la marche : vibrations du corps entier et étirements musculaires

Thomas Vialleron

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Analyse des effets d'une modification expérimentale  
de la capacité posturo-cinétique induite par deux  
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canique de la marche et de l'initiation de la marche :  
vibrations du corps entier et étirements musculaires  
*Effects of an experimental modification of the posturo-ki-  
netic capacity induced by two types of physical exercises on  
the biomechanical organization of walking and gait initia-  
tion: whole-body vibrations and muscle stretching*

## Thèse de doctorat de l'université Paris-Saclay

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Thèse présentée et soutenue à Paris-Saclay,  
le 17/03/2021, par

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A Marine et à notre fille, Louise



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# Préambule

## Préambule

Le contrôle de l'équilibre est une exigence capitale qui sous-tend l'exécution de chacune de nos actions motrices. Nous vivons en effet dans un environnement gravitaire qui tend à nous attirer en permanence vers le sol. De plus, chacun de nos mouvements génère, en vertu du principe de l'action et de la réaction, des forces internes qui sont transmises quasi-instantanément le long du corps par le biais des os et des tissus mous. A ces différentes sources de perturbation, s'ajoutent celles, beaucoup plus faibles, induites par les phénomènes végétatifs tels que la respiration, les battements cardiaques, le péristaltisme etc.

La question de savoir comment le système nerveux central parvient à contrôler l'équilibre malgré ces différentes sources de perturbations n'est pas nouvelle et a été abordée par de nombreux (neuro)physiologistes du 20<sup>ème</sup> siècle et de la fin du 19<sup>ème</sup> siècle, tels que Babinski (1899), Hess (1943) ou encore André-Thomas (1940) pour ne citer qu'eux, mais le développement récent des techniques d'analyse du mouvement et l'émergence de nouveaux concepts d'organisation neurophysiologique et biomécanique du mouvement ont permis d'appréhender dans toute sa complexité ce contrôle chez le sujet sain, et ses dysfonctionnements chez le sujet pathologique.

Un des paradigmes expérimentaux aujourd'hui classiquement utilisé pour investiguer ce contrôle lors de la production de tâches complexes dynamiques est celui de l'« initiation de la marche ». C'est ce paradigme qui est utilisé dans ce mémoire de thèse. L'initiation de la marche est une tâche fonctionnelle qui correspond à la période transitoire entre deux états stables, celui de la station debout et celui de la marche stationnaire. Cette tâche, que nous effectuons quotidiennement, est un challenge pour le système de contrôle postural. Pour effectuer le premier pas de la marche, nous devons en effet passer d'une situation stable et statique sur deux jambes à une situation instable et dynamique sur une seule jambe. Si le centre des masses n'est pas repositionné vers la jambe portante au moment du décollement du pied oscillant, celui-ci aura naturellement tendance à chuter latéralement du côté de la jambe

oscillante sous l'effet de la gravité. De plus, le déplacement vers l'avant de la jambe oscillante génère, en vertu du principe de l'action et de la réaction, des forces internes appliquées au niveau de l'articulation coxo-fémoral. Ces forces, initialement dirigées vers l'arrière et vers le bas, sont perturbatrices de l'équilibre. Enfin, au moment de l'impact du pied au sol, des forces de réaction sont produites et transmises instantanément de façon ascendante le long des segments corporels jusqu'à la tête. Le système postural doit être en mesure de s'opposer à l'ensemble de ces perturbations via le développement de phénomènes dynamiques contre-perturbateurs, pour assurer une progression stable du corps. Ces phénomènes correspondent aux « ajustements posturaux ».

L'on sait aujourd'hui que le développement adéquat de ces ajustements posturaux nécessite l'intégration de nombreuses informations sensorielles (cutanées, proprioceptives, vestibulaires et visuelles), et mobilise de façon hautement coordonnée une part étendue de la musculature striée squelettique. L'altération de tout ou partie des voies nerveuses et/ou des structures sensorimotrices impliquées dans ce contrôle, comme c'est le cas chez la personne âgée, suite au vieillissement physiologique ou pathologique (comme chez les patients ayant subi un AVC ou atteints de la maladie de Parkinson) entraîne irrémédiablement un risque de chute accru. Ainsi, il a été montré que l'altération de la capacité des sujets à transférer leur poids du corps vers la jambe portante au moment du décollement du pied oscillant lors de l'initiation de la marche (ce qui correspond à un ajustement postural) représente un risque de chute majeur chez la personne âgée (Robinovitch et al. 2013).

Selon l'organisation mondiale de la santé (World Health Organization 2008), les chutes constituent la deuxième cause de mortalité par accident. Environ un tiers des plus de 65 ans vivant à domicile chutent chaque année, près d'une sur deux passé 80 ans. La fréquence des chutes augmente avec l'âge et le niveau de fragilité. La chute peut aussi entraîner un syndrome post-chute, dans 1 cas sur 5 environ (Morisod and Coutaz 2007), responsable d'une perte d'autonomie et de troubles psychologiques favorisant une entrée précoce dans la dépendance. Partout dans le monde, les

personnes de plus de 60 ans sont de plus en plus nombreuses. Les plus de 80 ans représenteront 20% des personnes âgées en 2050. La prévention des chutes est donc un enjeu majeur de notre société car les impacts socio-économiques et sur la santé sont considérables pour les personnes, les familles et la communauté.

Pour se prémunir de ses conséquences qui peuvent être dramatiques, il est important, entre autres, de changer nos comportements. Adopter une vie saine avec de l'activité physique est un moyen efficace de vieillir en bonne santé et d'éviter les chutes. La pratique d'exercices de renforcement musculaire, d'équilibre, de flexibilité, ou simplement de marche, sont facilement réalisables, peu coûteux et ont montré leur efficacité pour réduire les chutes chez les personnes âgées. Pour que de telles interventions réussissent, il est crucial de changer les habitudes et les croyances des personnes âgées et des praticiens. Le succès des programmes repose souvent sur l'assiduité aux séances et leur continuité. Pour que cela fonctionne, il faut que le programme corresponde aux capacités physiques, mentales et sociales des personnes et que les changements soient perceptibles et bénéfiques pour elles, supérieurs aux obstacles et aux efforts nécessaires pour s'entraîner.

Cependant, les connaissances actuelles sur les effets de la pratique physique sur le contrôle postural et la marche restent encore limitées.

Aussi, ce travail de thèse se propose d'étudier les effets de deux types d'exercices physiques sur les processus locomoteurs de la marche et l'initiation de la marche dans différents types de population (saine ou pathologique). Il s'agit des vibrations corps entier (VCE) et des étirements musculaires, qui correspondent tous deux à des exercices très répandus en rééducation, et susceptibles de modifier la « capacité posturo-cinétique » des participants en augmentant potentiellement la raideur des membres inférieurs et la mobilité articulaire, respectivement.

Cette thèse s'articule autour de 4 chapitres.

Dans le **premier chapitre**, nous aborderons les principaux concepts biomécaniques et neurophysiologiques qui composent le cadre théorique de ce travail de recherche. Plus spécifiquement, nous rappellerons les lois fondamentales de la mécanique qui sous-tendent l'organisation du mouvement, puis aborderons les concepts d'ajustements posturaux et de capacité posturo-cinétique. Nous aborderons ensuite l'organisation neurophysiologique des ajustements posturaux, puis présenterons le paradigme de l'initiation de la marche utilisé lors de nos études expérimentales. Enfin, nous traiterons des répercussions du vieillissement et de la maladie de Parkinson sur le contrôle postural lors de cette tâche locomotrice, ainsi que des effets de l'exercice physique sur l'organisation des ajustements posturaux. A noter qu'il est actuellement bien reconnu que ce processus est d'un intérêt particulier pour l'étude des effets du vieillissement et de la maladie de Parkinson sur le contrôle postural (e.g. Delval et al. 2014).

La problématique de ce mémoire de thèse sera présentée dans le **second chapitre**.

Dans le **troisième chapitre**, nous reporterons nos travaux. Ceux-ci sont présentés sous forme d'articles publiés et organisés en deux grandes parties. Dans la première partie, nous présenterons (1) une revue systématique de la littérature avec méta-analyse sur les effets à long terme des VCE sur la marche dans différents types de population (saines ou pathologiques) puis, (2) une étude expérimentale sur l'impact immédiat des VCE sur les mécanismes du contrôle postural lors du paradigme de l'initiation de la marche chez le jeune adulte et la personne âgée. De la même façon, dans la deuxième partie, nous présenterons (1) une revue systématique de la littérature avec méta-analyse sur les effets à long terme des étirements sur la marche dans différents types de population (saines ou pathologiques) puis, (2) une étude expérimentale sur les effets immédiats d'une séance d'étirement du triceps sural sur les mécanismes du contrôle postural lors de l'initiation de la marche chez la personne âgée et le patient atteint de la maladie de Parkinson.



Enfin, dans le **quatrième chapitre**, nous apporterons nos conclusions avant d'évoquer des perspectives cliniques et de recherche.

Les travaux présentés dans ce mémoire de thèse permettront de mieux comprendre les effets immédiats des VCE et des étirements musculaires sur les mécanismes du contrôle postural, ainsi que leurs effets à long terme sur la marche, le risque de chute et l'autonomie des personnes âgées (avec et sans pathologie associée). Nos données contribueront ainsi à améliorer les recommandations et les décisions thérapeutiques des cliniciens en charge de ces populations.

# **Chapitre 1 : Cadre théorique**

# 1. Cadre théorique

## 1.1. Principes fondamentaux de la mécanique appliqués à l'étude du contrôle postural et du mouvement

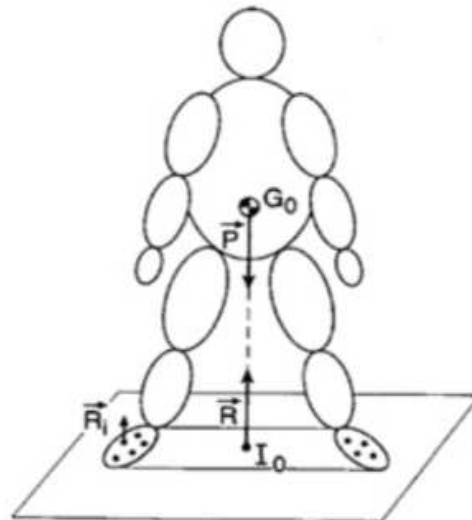
### 1.1.1. Posture et équilibre statistique

La position érigée est la posture qui caractérise l'espèce humaine. Elle est fondamentale car essentielle pour réaliser la plupart de nos activités quotidiennes. La posture se définit classiquement comme la position dans l'espace des segments corporels les uns par rapport aux autres à un moment donné (Massion 1992). À chaque posture correspond un état d'équilibre au sens physique du terme.

D'un point de vue mécanique, l'équilibre statique d'un corps par rapport à un référentiel Galiléen est satisfait si et seulement si les deux conditions qui régissent le principe fondamental de la statique sont respectées. La première, issue de la seconde loi de Newton, implique que la résultante des forces extérieures appliquées au système soit nulle. La seconde, provenant de la loi d'Euler relative à la dynamique du mouvement en rotation, implique que la somme des moments résultants de ces mêmes forces soit nulle en tous points.

Lorsqu'un individu se tient en posture debout sans autre contact avec l'environnement que ses soles plantaires, il n'est soumis qu'à deux forces externes. La première, qui est liée à l'action de la gravité terrestre, correspond au poids. Cette force s'applique au niveau du centre des masses global du sujet (qui correspond au barycentre du poids de chacun des segments corporels) et tend à le plaquer au sol. La seconde, qui est liée au support sur lequel se tient le sujet, correspond à la force de réaction du sol. Cette force s'applique au niveau du centre des pressions, qui correspond au barycentre des forces de réaction du sol. Ainsi, pour que la posture debout puisse être maintenue, les conditions de l'équilibre énoncées ci-dessus

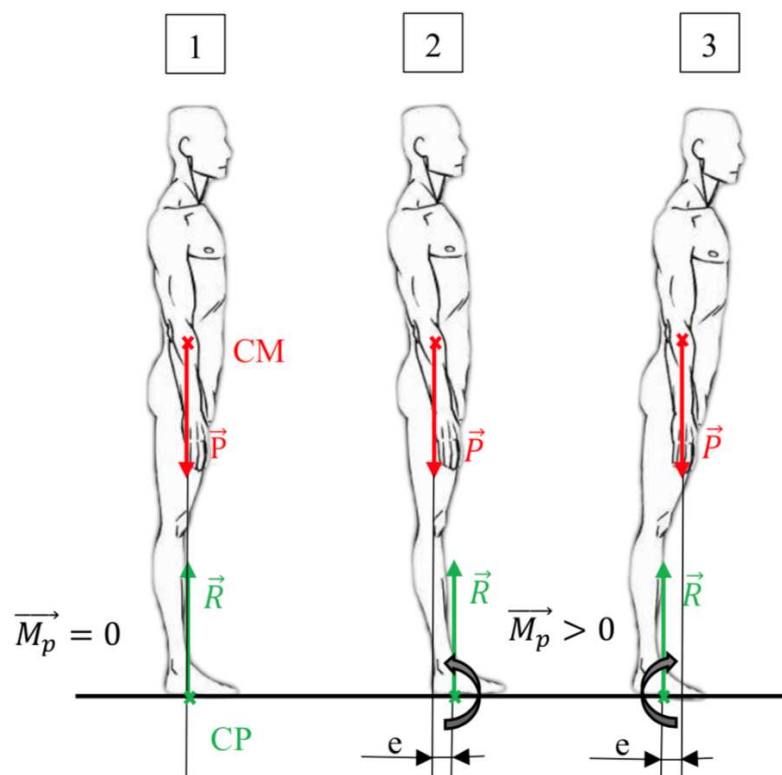
impliquent 1) que la résultante des forces de réaction du sol et le poids du sujet soient de même intensité et de sens opposé et 2) que le centre des pressions soit situé sur la verticale passant par le centre des masses (Figure 1). Par extension, il est généralement considéré que l'équilibre est maintenu lorsque que le centre des masses se projette dans la zone délimitée par la base de support.



**Figure 1 : Équilibre statique en position debout.** Lorsque le sujet est debout au repos, il est soumis à la force de gravité ( $\vec{P}$ ) et aux forces de réaction du sol ( $\vec{R}$ ). La première s'applique au centre des masses ( $G_0$ ) et les autres au centre des pressions ( $I_0$ ). Le centre des pressions est le barycentre de toutes les forces de réaction qui s'appliquent sous les pieds du sujet ( $\vec{R}_i$ ). L'équilibre posturale est assuré car les deux vecteurs de force,  $\vec{P}$  et  $\vec{R}$ , sont alignés sur la verticale gravitaire (ligne pointillée). D'après Bouisset and Do (2008).

Toutefois, le corps est formé d'un ensemble de segments corporels (tête, tronc, bras, cuisses, jambes, pieds) qui se mobilisent autour des articulations. Il constitue donc un système déformable. Par conséquent, pour que l'équilibre soit maintenu, il faut, en plus des conditions précédentes, que le centre des masses de chacun de ces segments et leur centre de rotation soient alignés sur la verticale gravitaire ce qui, en pratique, n'est pas réalisable notamment du fait de la forme des os (Bouisset and Maton 1996). Notons par exemple, que la projection du centre des masses du corps se situe légèrement en avant de l'articulation de la cheville. Cet écart par rapport au pivot de

l'articulation crée un bras de levier et engendre mécaniquement un moment de force autour de la cheville qui tend à faire basculer le corps en avant sous l'effet de la gravité. Ce phénomène implique que le maintien de la posture debout n'est possible que si les moments des forces gravitaires sont compensés par les forces internes (musculaires et ligamentaires) qui agissent pour créer des moments de forces de sens opposés autour de l'articulation de la cheville (Figure 2). Encore une fois, l'équilibre n'est obtenu que si les moments de forces gravitaires et musculaires s'annulent.



**Figure 2 : Rétablissement de l'équilibre statique en position debout.** Sur la position 1, le centre des masses (CM) et le centre des pressions (CP) sont sur la même ligne gravitaire. Il n'y a aucun moment induit. Sur les positions 2 et 3, la projection du CM n'est plus confondue avec le CP. La distance entre le CM et le CP ( $e$ ) induit un moment autour de la cheville sous l'effet du poids du corps ( $\vec{P}$ ). Le corps bascule vers l'arrière (2) ou vers l'avant (3) et les muscles du sujet vont devoir se contracter pour contrecarrer ces déséquilibres. Ces ajustements doivent être rapides car plus distance entre le CM et le CP ( $e$ ) augmente, plus le CM est accéléré en dehors de la base de support. Les déplacements sont ici décrits d'avant en arrière (sur l'axe antéropostérieur) mais se retrouvent aussi sur les côtés, de gauche à droite (sur l'axe médiolatéral). D'après Winter et al. (1990).

Il est aujourd'hui largement connu que l'être humain en posture debout n'est jamais dans un état d'équilibre statique parfait, en témoigne l'existence d'oscillations permanentes du centre des masses et du centre des pressions, ou « oscillations posturales » (e.g. Romberg 1953 ; Bouisset and Duchêne 1994). Ces oscillations ont pour origines, entre autres, l'activité des pompes cardiaques et respiratoires, le péristaltisme, les fluctuations de la commande motoneuronale etc. (e.g. Bouisset and Duchêne 1994 ; Hamaoui et al. 2002). Ces oscillations posturales reflètent un phénomène « global », i.e. résultant à la fois de la perturbation à la posture (induite par les phénomènes décrits ci-dessus) et de la contre-perturbation qui lui est associée. Nous reviendrons sur ces notions de « perturbation » et de « contre-perturbation » (partielle), qui sont au cœur du concept de « capacité posturo-cinétique », plus loin dans l'exposé.

Les oscillations posturales sont classiquement utilisées dans la littérature pour quantifier le degré de stabilité des individus dans une posture donnée (e.g. Schieppati et al. 1994 ; Hamaoui et al. 2004 ; Vuillerme and Nafati 2007 ; Schniepp et al. 2013 ; Debove et al. 2017). Cette stabilité témoigne de l'efficacité de l'ensemble du système postural. De très nombreux paramètres de ces oscillations ont été considérés dans la littérature pour quantifier cette stabilité posturale. Un certain nombre d'entre eux sont présentés dans l'article de Hansen et al. (2017).

### **1.1.2. Équilibre dynamique**

Comme il a été dit plus haut, dans le cas du maintien postural statique, l'équilibre est classiquement considéré comme « stable » lorsque la projection au sol du centre des masses reste à l'intérieur de la base de support (e.g. Borelli 1960 ; Kuo 1995). En condition dynamique, Pai et collaborateurs (e.g. Pai and Patton 1997), ainsi que Hof et collaborateurs (e.g. Hof et al. 2005), ont montré que cette condition n'était plus suffisante pour qu'un état d'équilibre stable puisse être atteint. Ainsi, Hof et al. (2005), dans leur article intitulé « *the condition for dynamic stability* », ont élaboré le

concept de « centre des masses extrapolé » pour tenir compte du fait que la vitesse du centre des masses, négligeable en condition de maintien postural statique, intervenait de façon non négligeable en condition dynamique. Dans leur modèle, le centre des masses extrapolé est une variable qui dépend à la fois de la position du centre des masses et de sa vitesse de déplacement. Il s'agit donc d'une variable « synthétique ». Le terme « extrapolé » exprime le fait que la trajectoire du centre des masses est « extrapolée » dans la direction de sa vitesse. Dans ce modèle, le maintien de l'équilibre dynamique implique que le centre des masses extrapolé (et non plus le « centre des masses ») reste contenu dans la base de support. Dans le cas contraire, il faudra que le sujet entreprenne des mouvements de rattrapage pour ramener le centre des masses extrapolé à l'intérieur de la base de support et ne pas chuter. Cette condition peut être considérée comme une extension des conditions de stabilité décrites plus haut pour l'équilibre statique.

Au final, si posture et équilibre constituent deux aspects intimement liés d'un même phénomène, leurs relations ne sont toutefois pas biunivoques : du fait de la structure du corps humain, une même projection du centre des masses (ou du centre des masses extrapolé) à l'intérieur de la base de support peut résulter d'une infinité de configurations segmentaires (i.e. de postures). La position relative des segments corporels et le maintien de l'équilibre ne procèdent donc pas nécessairement des mêmes mécanismes, comme l'ont proposé Do et al. (1991).

### **1.1.3. Approche biomécanique du mouvement**

Pour se mouvoir et se déplacer, un individu doit nécessairement prendre appui sur un support du milieu extérieur afin de créer des forces de réaction qui permettront la rupture de l'équilibre postural initial. Cette nécessité résulte des lois de la mécanique newtonienne : la loi de la résultante dynamique (seconde loi de Newton) et la loi du moment cinétique (loi d'Euler). Ces lois permettent de relier à chaque instant forces et

mouvement. De ces deux lois découlent au moins deux principes fondamentaux qui sont le *principe d'inertie* et le *principe de l'action et de la réaction*.

Selon le *principe d'inertie*, si le corps d'un individu est au repos par rapport au milieu extérieur, seule une force extérieure peut le mettre en mouvement (Gray 1968). Ces forces extérieures sont principalement la force de gravité et la force de réaction du sol (ainsi que les forces de frottements de l'air, de l'eau, etc., considérées ici comme négligeables). Pour se déplacer par ses propres efforts, un individu doit donc faire naître une force (ou un moment) de son environnement extérieur. Le système nerveux central ne pouvant agir directement sur la gravité, le déplacement du corps ne peut se réaliser que par une action sur les forces de réaction du sol. Lors d'un mouvement volontaire, l'activité musculaire permet de faire naître ces forces de réaction sur les surfaces d'appui offertes par l'environnement extérieur. Le système nerveux doit contrôler ces forces de façon à ce que le corps se déplace dans la direction voulue.

Le fait qu'il y ait des réactions aux actions des muscles implique également que les mouvements segmentaires provoquent des forces de réaction autour des articulations mobilisées. Comme dit précédemment, le corps humain est composé d'un ensemble de solides liés entre eux, qui se déplacent les uns par rapport aux autres en formant des chaînes articulaires qui interagissent entre elles. Ainsi, chaque action mécanique qui s'exerce sur un point du corps est transmise sans délai à toutes les autres parties du corps. Les forces développées durant le mouvement sont transmises des segments mobilisés volontairement vers tous les autres, jusqu'aux surfaces de contact au sol d'où naissent les forces de réaction. En d'autres termes, là où les forces musculaires provoquent un mouvement segmentaire, le reste du corps aura tendance à se déplacer dans la direction opposée selon le *principe d'action-réaction*. Ce principe implique que tout mouvement a un effet déstabilisant sur l'ensemble du corps, et donc qu'une « contre-perturbation » doit nécessairement être développée pour que le corps puisse être maintenu en équilibre. Cette « contre-perturbation » revêt la forme de phénomènes dynamiques survenant au niveau des segments posturaux, et connus sous le terme d'« ajustements posturaux ».



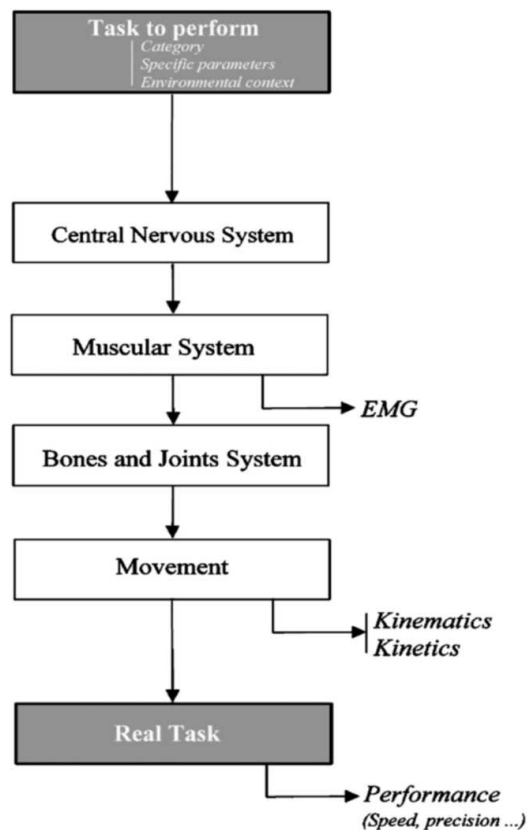
Dans le cadre de cette thèse, c'est au mouvement volontaire que nous nous intéressons, et à la capacité du système nerveux central de s'opposer à la perturbation posturale induite par celui-ci. Aussi, dans le paragraphe suivant, nous aborderons le concept de « mouvement volontaire » et d'« ajustement postural ».

#### **1.1.4. Mouvement volontaire et perturbation de l'équilibre**

##### **1.1.4.1. Le mouvement volontaire**

Le mouvement est dit « volontaire » lorsqu'il répond à l'intention de réaliser une tâche déterminée (Bouisset and Do 2008). On distingue la tâche *prescrite*, qui est celle que l'on désire réaliser, de la tâche que l'on réalise vraiment et qui est la tâche *réelle* (Figure 3). Entre les deux, le système nerveux central planifie des stratégies motrices adaptées et émet des ordres au muscles qui mobilisent les segments corporels autour des articulations. La tâche prescrite se caractérise par le type d'activité à laquelle elle appartient (pointer, lancer, courir, marcher etc.), par les paramètres spécifiques qui sont demandés (vitesse, amplitude, orientation, précision), et enfin par son environnement (stabilité du sol, résistances à l'avancement, de l'eau ou du vent, luminosité, bruits etc.). De ce point de vue, la tâche prescrite représente une source d'informations multiples pour les systèmes sensoriel et moteur. De son côté, la tâche réelle est le résultat de la capacité du sujet à mobiliser ses ressources physiques et mentales pour répondre au mieux aux contraintes de la tâche prescrite. L'efficacité de la tâche réelle est évaluée avec la performance. La performance est mesurée par des critères qui permettent d'apprécier l'adéquation de la tâche réalisée à la tâche souhaitée par rapport à la vitesse, précision, amplitude etc. Le fait que la tâche réalisée réponde aux impératifs prescrits, c'est-à-dire que la tâche réelle soit conforme avec la tâche prescrite, constitue la preuve de l'efficacité du mouvement. En d'autres termes, l'exécution du mouvement et la réalisation de la tâche constituent deux aspects différents mais intimement liés du même phénomène, mais le mouvement n'est que le moyen de réaliser la tâche. La

réalisation du mouvement demande la mobilisation de certains segments corporels (correspondant aux segments *focaux*) et l'immobilisation d'autres segments (correspondant aux segments *posturaux*).



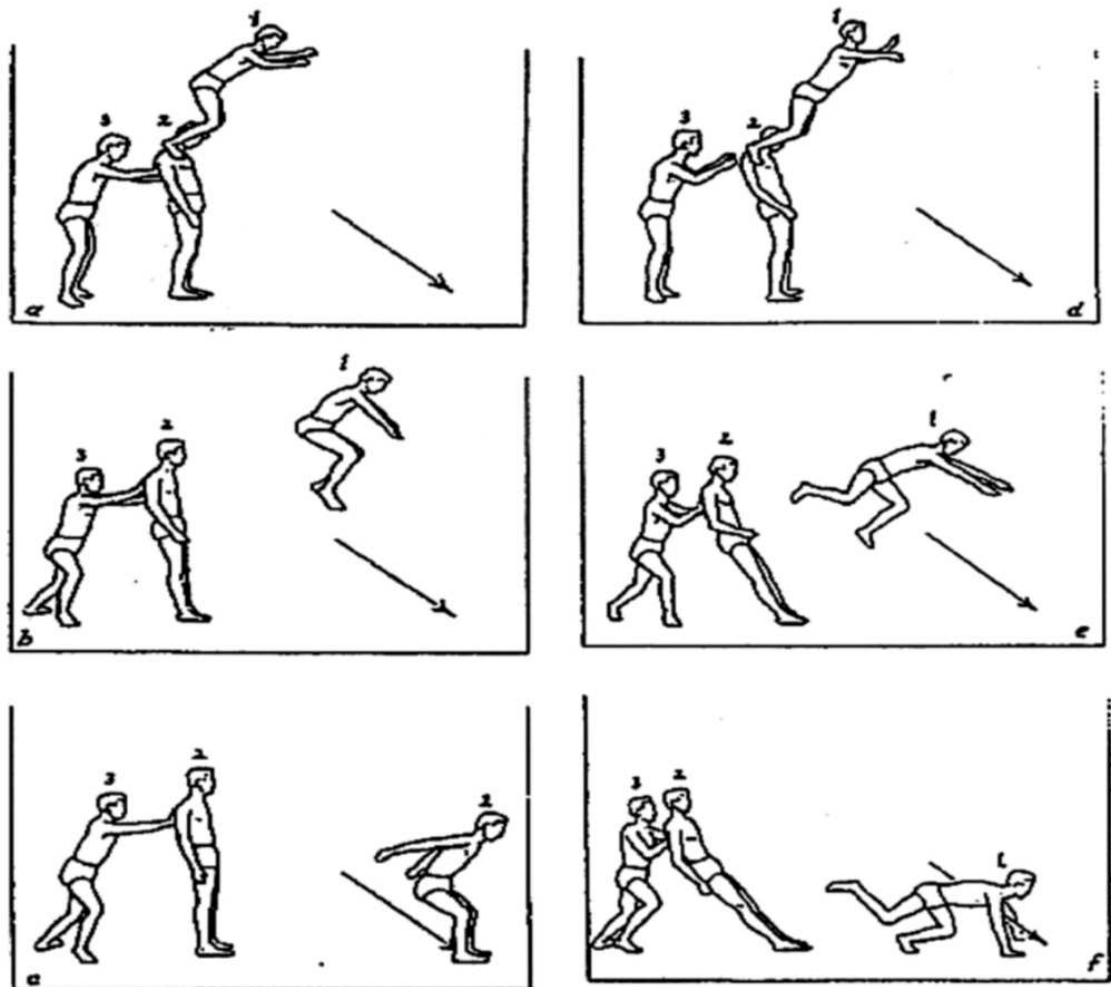
**Figure 3 : De la tâche prescrite à la tâche réelle.** La tâche prescrite (*task to perform*) est la tâche que le sujet tente de réaliser. Elle se caractérise par 3 facteurs : sa catégorie, ses paramètres spécifiques et son contexte environnemental (*category, specific parameters, environmental context*). Ces facteurs sont pris en compte par le système nerveux central (*central nervous system*) lors de la programmation motrice, puis celui-ci envoie des commandes pour activer les muscles (*muscular system*). Les muscles mobilisent les os et les articulations (*muscular and joints systems*), ce qui crée le mouvement. La tâche réelle est la tâche exécutée par le sujet (*real task*). L'activité musculaire peut être mesurée grâce à de l'électromyographie (EMG), les mouvements peuvent être quantifiés grâce aux lois de la dynamique (*kinetics*) et à la cinématique de leurs paramètres spatio-temporels (*kinematics*). L'efficacité du mouvement est évaluée au travers de la performance par la vitesse (*speed*), la précision (*precision*) etc... D'après Bouisset and Do (2008).

#### 1.1.4.2. « Dualité » entre la posture et le mouvement

Les travaux du neurologue André-thomas et du physiologiste Walter R Hess à la fin de la première moitié du XXe siècle ont montré que le système moteur s'organise autour d'une distinction entre la posture et le mouvement. Cette conception est aujourd'hui confortée par des études récentes montrant l'existence d'un contrôle distinct de la posture et du mouvement au niveau du cortex moteur primaire chez le singe (e.g. Kurtzer et al. 2005).

Dans un modèle imagé (Figure 4), Hess (1943) distinguait dans l'organisation du mouvement volontaire trois composantes, symbolisées chacune par un personnage : un sauteur, qui se tient debout sur les épaules d'un porteur, et un pareur en support du porteur. Deux notions se dégagent de ce modèle : premièrement, le mouvement volontaire (symbolisé par le bond du sauteur) est perturbateur de l'équilibre (symbolisé par le porteur), l'action du sauteur risquant de faire basculer en arrière le porteur. Deuxièmement, il y a une dichotomie des réactions d'équilibration. On peut distinguer une composante « statique » incarnée par le porteur et une composante « dynamique » incarnée par le pareur qui s'oppose à la perturbation du mouvement volontaire (le sauteur). Ce modèle de l'organisation du mouvement volontaire illustre ainsi les notions de « support postural », de « mouvement focal » et d'« ajustement postural ». Il met également en avant la nécessité de coordonner ces différentes composantes du mouvement volontaire afin que la tâche puisse être réalisée efficacement.

Au final, le mouvement volontaire est donc soumis à deux contraintes. Déplacer un ou plusieurs segments du corps vers un but (le mouvement focal) et stabiliser d'autres segments (la chaîne posturale) afin de maintenir la posture et l'équilibre. Comme souligné par Agid (1991), l'une des principales tâches du système nerveux central est de coordonner la posture et le mouvement.



**Figure 4 : Le modèle de Hess.** Ce modèle inclue trois composantes, chacune représentée par un personnage : la sauteur (1), qui se tient sur les épaules du porteur (2), et le pareur (3) qui sécurise le porteur au moment du saut. Le sauteur représente le mouvement volontaire, le porteur et le pareur représentent respectivement l'équilibration statique et dynamique. A gauche (images a, b et c), la tâche est exécutée efficacement. Grâce à un contrôle postural optimal, l'équilibre dynamique est maintenu, ainsi que la récupération de l'équilibre statique à la fin du saut. A droite (images d, e et f), la tâche n'est pas réalisée efficacement, le sujet tombe à la fin du saut. La stabilisation posturale n'est pas suffisamment efficace pour contrebalancer l'effet déstabilisateur du saut. D'après Bouisset and Do (2008).

#### 1.1.4.3. Mouvement volontaire et activité posturale

Le modèle théorique de Hess décrit ci-dessus illustre le fait que le mouvement volontaire ne peut être effectué efficacement (la tâche réalisée correspondant alors à la tâche prescrite) sans une activité posturale adaptée. Cette activité posturale a depuis

été étudiée de façon expérimentale et observée au travers de la production de nombreux mouvements volontaires, tels que :

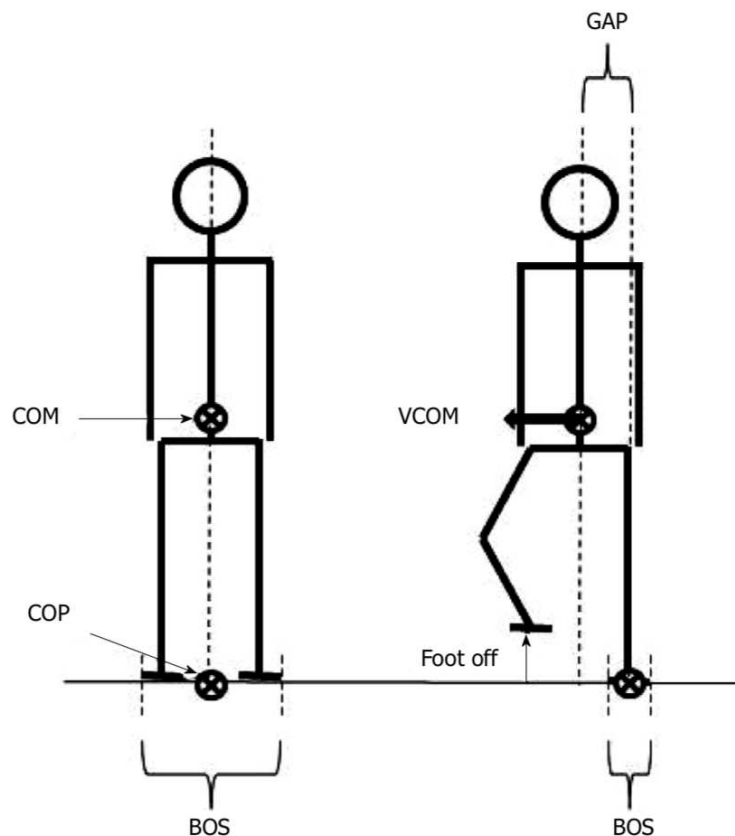
- des mouvements segmentaires sans modification de la base de support, comme l'élévation antérieure du membre supérieur (e.g. Bouisset and Zattara 1981, 1987 ; Dick et al. 1986 ; Lee et al. 1987), l'abduction du membre supérieur (e.g. Vernazza et al. 1996), la flexion du coude (e.g. Friedli et al. 1988 ; Benvenuti et al. 1995), la flexion des doigts (Bolzoni et al. 2018), des poussées et tirages d'une poignée (e.g. Cordo and Nashner 1982) et la flexion du tronc (e.g. Oddsson and Thorstensson 1987 ; Crenna et al. 1987 ; Pedotti et al. 1989) ;
- Des mouvements segmentaires avec modification de la base de support, comme la flexion d'un membre inférieur en position debout avec passage en appui unipodal (Rogers and Pai 1990 ; Do et al. 1991 ; Nouillot et al. 1992 ; Yiou et al. 2012 a, b ; Gendre et al. 2016) ;
- Des mouvements impliquant un déplacement de l'ensemble du corps avec réduction de la base de support comme la montée sur la pointe des pieds ou rouler sur les talons (Nardone and Schieppati 1988) ;
- Et enfin, des mouvements locomoteurs impliquant une progression de l'ensemble du corps, comme la locomotion et l'initiation de la marche (Brenière et al. 1987 ; Caderby et al. 2014 ; Honeine et al. 2016 ; Lepers and Brenière 1995 ; Mann et al. 1979 ; Winter 1995 ; Yiou et al. 2016b).

Selon les lois de la mécanique énoncées précédemment, chacun de ces mouvements constitue une perturbation transitoire de la posture et de l'équilibre pour deux raisons majeures. Tout d'abord, parce que l'accélération du segment focal génère, en vertu du *principe de l'action et de la réaction*, des forces internes qui sont appliquées au niveau de l'articulation proximale. Ces forces internes sont transmises quasi-instantanément le long de la chaîne posturale jusqu'aux surfaces d'appui et sont responsables d'une variation des forces de réaction agissant au niveau de ces surfaces. Celles-ci étant des forces externes au corps, leur variation constitue une source de

perturbation de l'équilibre postural préétabli. Deuxièmement, tout mouvement segmentaire modifie la position du centre des masses de l'ensemble du corps, impliquant de nouvelles conditions posturales afin que le centre des masses puisse à nouveau être positionné au-dessus du centre des pressions.

Pour reprendre le paradigme de l'élévation du membre supérieur classiquement étudié dans la littérature (e.g. Belen'kiï et al. 1967; Bouisset and Zattara 1981, 1987 ; Dick et al. 1986 ; Lee et al. 1987), l'analyse par modélisation des forces internes d'origine musculaire générées au niveau de l'épaule (segment proximal du membre focal) montre que celles-ci ont initialement une résultante dirigée vers l'arrière et vers le bas (Bouisset and Zattara 1981, 1987). Ces forces internes engendrent des forces de réaction au niveau de l'épaule et jusqu'au niveau des surfaces d'appui qui ont tendance à déséquilibrer le corps dans la même direction. Ces forces s'inversent lorsque le mouvement focal décélère et se stabilise. A ce moment-là, la nouvelle position du bras vers le haut et vers l'avant induit un déplacement de la position initiale du centre des masses vers l'avant.

Dans le cas des tâches locomotrices comme l'initiation de la marche, le fait de lever le pied du sol entraîne une diminution drastique de la base de support, le sujet passant alors d'un appui bipodal à un appui unipodal (Figure 5). Sans action entreprise pour repositionner le centre des masses au-dessus de la nouvelle base de support, le sujet chuterait rapidement du côté de la jambe oscillante au cours de la phase d'exécution du pas (Brenière et al. 1987 ; Yiou et al. 2016b, 2017).



**Figure 5 : représentation des déséquilibres lors de l'initiation de la marche.** A gauche, le centre des masses (COM) est projeté dans la base de support (BOS), sur la même ligne que le centre des pressions (COP), le corps est à l'équilibre. A droite, lorsque le sujet lève le pied (*Foot off*), la base de support est considérablement réduite. Le centre des masses et le centre des pressions ne sont plus alignés, ce qui entraîne la chute du centre des masses vers la jambe oscillante. Un écart (*Gap*) entre le centre des masses et le centre des pressions se crée au fur et à mesure de la chute. Plus cet écart augmente, plus la vitesse du centre des masses (VCOM) s'accélère. D'après Yiou et al. (2017).

Le maintien de l'équilibre postural nécessite donc le développement de phénomènes dynamiques qui vont s'opposer aux différentes sources de perturbations induites par le mouvement. On parle d' « ajustements posturaux » pour faire référence à ces phénomènes dynamiques contre-perturbateurs. Les caractéristiques de ces ajustements posturaux sont décrites aux paragraphes suivants.

#### 1.1.4.4. Les ajustements posturaux

L'on sait aujourd'hui que les ajustements posturaux peuvent précéder, accompagner, et suivre le mouvement volontaire (Gahery 1987 ; Bouisset and Do 2008).

Les ajustements posturaux *préparatoires* permettent le positionnement du corps adapté à la tâche à réaliser. Ces ajustements apparaissent précocement avant le mouvement volontaire, sans modification de la posture initiale. Leur but est de placer le corps dans une position favorable à l'exécution de la tâche. Pour revenir aux exemples cités précédemment, lors de l'élévation du membre supérieur (Bouisset and Zattara 1981, 1987) ou de l'initiation d'un pas rapide (Lepers and Brenière 1995), le centre des masses est porté préalablement vers l'avant pour, respectivement, contrecarrer les forces de réactions orientées vers le l'arrière et permettre un départ rapide.

Les ajustements posturaux *d'accompagnement* surviennent en même temps que la réalisation du mouvement focal ou peuvent précéder son initiation. Ils correspondent alors aux « ajustements posturaux anticipateurs » (APA), et aux « ajustements posturaux simultanés » (Fourcade et al. 2014, 2016 ; Fourcade and Memari 2015). Les APA font l'objet d'une attention particulière dans le chapitre 1.1.5.

Les ajustements posturaux *correcteurs*, parfois nommés *consécutifs*, correspondent par définition aux phénomènes posturaux survenant après le mouvement volontaire. Ils ont été observés lors du maintien d'un équilibre instable (Day et al. 1993), lors du retour à l'équilibre après l'exécution du mouvement focal (Bouisset and Zattara 1987 ; Le Bozec et al. 2008 ; Memari et al. 2013), ou encore après la survenue d'un évènement extérieur venant perturber l'équilibre, tels que le choc induit par un pendule percutant de façon imprévisible les épaules du sujet, le déplacement brusque de la surface d'appui etc. (e.g. Horak and Nashner 1986 ; Aruin et al. 2015, 2017 ; Kanekar 2015).



Parmi ces différents types d'ajustements posturaux, ceux qui ont fait l'objet du plus grand nombre d'études sont incontestablement les APA. Ces APA ont en effet un statut particulier : survenant par définition avant la production du mouvement volontaire, ils ne peuvent pas être déclenchés de façon réflexive par celui-ci et résultent donc d'une programmation par le système nerveux central. Leur étude permet donc d'investiguer de façon indirecte la programmation de la commande posturale associée au mouvement volontaire. Le fait que des ajustements posturaux soient également contemporains du mouvement n'exclue pas le fait qu'ils soient également programmés, mais leur mise en évidence nécessite le développement de modèles biomécaniques permettant de distinguer la composante focale de la composante posturale dans les signaux recueillis à la périphérie, ce qui rend leur analyse plus complexe (cf. Memari et al. 2020). Les caractéristiques principales des différents types d'ajustements posturaux sont décrites dans les paragraphes suivants.

### **1.1.5. Ajustements posturaux anticipateurs (APA)**

Les APA ont initialement été mis en évidence par Belen'kiï et al. (1967) lors du mouvement désormais devenu classique d'élévation du membre supérieur en posture debout (e.g. Bouisset and Zattara 1981, 1987 ; Dick et al. 1986 ; Lee et al. 1987). Plus spécifiquement, ces auteurs ont montré que les premiers muscles activés lors de l'élévation du membre supérieur n'étaient pas ceux situés autour de l'épaule mais ceux des membres inférieurs. D'autres auteurs (en particulier Bouisset et collaborateurs ; e.g. Bouisset and Zattara 1981, 1987 ; Zattara and Bouisset 1988) ont repris le même paradigme afin de décrire de façon plus complète les patrons EMG et dynamiques associés aux APA. Ainsi, il a été montré que l'activation du deltoïde antérieure (muscle responsable du mouvement d'élévation du bras) était systématiquement précédée de l'inhibition du soléaire ipsilatéral, suivie d'une activation du tenseur du fascia lata et du droit fémoral controlatéraux, du grand fessier et du demi-tendineux ipsilatéraux, et enfin des muscles érecteurs du rachis controlatéraux. Cette séquence est reproductible

et spécifique du mouvement. A cette séquence EMG « ascendante », est associée une séquence cinématique anticipatrice (Bouisset and Zattara 1988) débutant par une accélération du membre inférieur controlatéral, puis ipsilatéral, et se terminant par l'accélération du membre focal. L'existence de ces accélérations segmentaires qui surviennent au niveau de la chaîne posturale (de même qu'au niveau du centre des masses du corps, comme décrit au paragraphe suivant) témoignent du caractère dynamique de ces APA.

Les rôles fonctionnels de ces APA sont décrits dans les paragraphes suivants.

#### **1.1.5.1. Rôles fonctionnels des APA**

Le rôle fonctionnel des APA a été argumenté sur une base biomécanique (Bouisset and Zattara 1981, 1987 ; Cordo and Nashner 1982 ; Friedli et al. 1988 ; Bouisset and Do 2008). Si l'on reprend la tâche d'élévation du membre supérieur, les forces nécessaires à la réalisation du mouvement focal sont transmises le long de la chaîne posturale jusqu'aux surfaces d'appuis, selon le principe d'action-réaction. En théorie, ces forces sont initialement dirigées vers l'arrière et le bas, i.e. dans le sens opposé à l'élévation du bras, ce qui tend à déstabiliser le corps dans cette même direction. Bouisset et al. (1981, 1987) ont montré que des phénomènes dynamiques survenaient au niveau du centre des masses avant-même que cette perturbation ne soit opérante, et que ceux-ci (les APA donc) se caractérisaient par une accélération du centre des masses global vers l'avant et vers le haut, soit dans des directions diamétralement opposées aux forces réactionnelles générées par le mouvement du bras au niveau de l'épaule. C'est sur cette base que les auteurs ont proposé que les APA avaient pour fonction de créer par avance des forces inertielles au niveau du support postural permettant de contrebalancer la perturbation induite par le mouvement volontaire. Cette contre-perturbation posturale n'est cependant que partielle, en témoigne l'existence d'ajustements posturaux simultanés et consécutifs.

Les études reportées ci-dessus portaient sur l'organisation posturale associée à un mouvement volontaire réalisé sur une base posturale fixe. Le rôle contre-perturbateur des APA se retrouve également lors de mouvements induisant un changement de la base de support. Par exemple, il a été montré lors du mouvement de flexion de la cuisse à partir de la posture debout, que le décollement du pied était précédé par une dynamique posturale anticipatrice, caractérisée par un déplacement du centre des pressions vers la jambe oscillante (e.g. Rogers and Pai 1990 ; Do et al. 1991 ; Yiou et al. 2011b). Ce déplacement anticipateur du centre des pressions permettait le déplacement du centre des masses en sens opposé, i.e. vers la future jambe portante. Cette dynamique posturale anticipatrice permettait ainsi de repositionner le centre des masses à l'intérieur de la nouvelle base de support, et ainsi de pouvoir conserver l'équilibre pendant et à la fin du mouvement (Rogers and Pai 1990 ; Do et al. 1991 ; Yiou et al. 2011b). Nous retrouverons ce rôle stabilisateur des APA lors du processus d'initiation de la marche, qui sera décrit plus en détail dans la suite de l'exposé (chapitre 1.3).

Les APA n'ont cependant pas qu'un rôle exclusivement dévolu à la stabilisation posturale. Ils permettent également d'assister la performance motrice. En effet, dans les activités motrices nécessitant un déplacement du corps dans son ensemble, comme lors de l'initiation de la marche (Brenière et al. 1987), l'exécution d'un simple pas vers l'avant (Memari et al. 2013) ou encore la fente en escrime (Yiou and Do 2001), il a été montré que les APA permettaient de créer les conditions dynamiques nécessaires à la réalisation de l'objectif visé, e.g. atteindre une certaine longueur de pas/fente ou une vitesse de progression du corps. Ces APA se caractérisaient par un déplacement du centre des pressions vers l'arrière. Ce recul du centre des pressions permettait de créer un découplage avec le centre des masses dans le plan sagittal, et ainsi de propulser ce dernier vers l'avant. Ce type d'APA a également été mis en évidence lors du mouvement de se baisser pour ramasser un objet (ou « *whole-body reaching* » des anglo-saxons ; Stapley 1998, 1999). Là encore, nous reviendrons plus en détail sur ce processus lorsque nous aborderons l'initiation de la marche (chapitre 1.3).

### 1.1.5.2. Adaptabilité des APA

Une caractéristique majeure des APA est leur *adaptabilité*. Cette caractéristique est importante car nous évoluons dans un environnement intérieur et extérieur en perpétuel changement (cf. Yiou et al. 2012a). Ainsi, il a été montré à de nombreuses reprises dans la littérature que l'activité posturale anticipatrice était programmée de façon adaptative en fonction des paramètres du mouvement (vitesse, précision, force, charge, pression temporelle etc.), des conditions posturales (e.g. la stabilité de la posture initiale/finale, l'ajout d'une charge, la fatigue de la musculature posturale etc.), des conditions environnementales (e.g. la présence d'un obstacle à enjamber, la glissance des surfaces etc.). Les quelques exemples reportés ci-dessous permettent d'illustrer cette caractéristique des APA.

Lors du paradigme de l'élévation du membre supérieur, il a été montré que la vitesse maximale du mouvement focal, ainsi que l'inertie du bras, déterminaient les paramètres spatio-temporels des APA. A noter que plus la vitesse du mouvement et l'inertie du membre supérieur sont élevées, plus l'intensité de la perturbation posturale est importante. Dans ces conditions, il a été montré que la durée et l'amplitude des APA, et donc l'intensité de la contre-perturbation, augmentaient au prorata de ces deux paramètres du mouvement volontaire (Horak et al. 1984 ; Lee et al. 1987 ; Bouisset et al. 2000). Il est à noter que les APA ne sont présents qu'à partir d'un certain seuil de vitesse. D'autres auteurs ont mis en évidence l'effet de la précision sur les APA associés à une tâche de pointage vers une cible (e.g. Bonnetblanc et al. 2004 ; Caronni et al. 2013 ; Bruttini et al. 2016).

L'influence de la stabilité posturale initiale sur l'organisation des APA a été mise en évidence lors de mouvements mobilisant les membres supérieurs et inférieurs. Ainsi, Cordo and Nashner (1982) ont montré que lorsqu'un individu se tient debout dans une posture rendue très stable par l'ajout d'un appui thoracique, les APA associés à des mouvements rapides du bras (pousser et tirer une poignée) étaient atténués voire

supprimés. Ces auteurs ont proposé que lorsque la posture se trouve initialement dans un état très stable, les APA ne plus nécessaires. Les APA étant « coûteux » sur le plan énergétique, le système nerveux central en ferait l'économie. Cet effet de la stabilité initiale sur les APA a également été mise en évidence lors d'autres types de tâches motrices, telles que le pointage (Yiou et al. 2007), le lâcher d'une masse bras tendus (Aruin et al. 1998), ou encore la flexion de la cuisse (Do et al. 1991 ; Nouillot et al. 1992). Les APA étant des phénomènes dynamiques à part entière, si la base de support est réduite, la possibilité de générer des APA l'est aussi. De plus, dans ces conditions de grande instabilité, il a été proposé que les APA pouvaient constituer eux-mêmes une source de déséquilibre supplémentaire et étaient donc atténués.

Les effets d'une réduction de la base de support en posture initiale sur les APA ont également été étudiés lors de l'initiation de la marche (Couillandre et al. 2002 ; Couillandre and Brenière 2003). Dans ces études, les sujets étaient initialement positionnés sur la pointe des pieds, limitant ainsi la capacité de recul du centre des pressions au cours des APA. Couillandre et al. (2002) ont montré que, dans ces conditions, la diminution de l'amplitude du recul du centre des pressions était compensée par une augmentation de la durée des APA, permettant aux sujets d'atteindre un couple de déséquilibre entre le centre des pressions et le centre des masses au décollement du pied équivalent à celui obtenu en condition contrôle (pieds initialement entièrement en contact avec le sol). Il en résultait une performance motrice (vitesse maximale de progression du centre des masses et longueur de pas) inchangée.

De même, il a été montré que le port d'une ceinture lestée de 15% du poids du corps (charge équitablement répartie à hauteur du centre des masses des sujets), entraînait des modifications adaptatives de l'amplitude des APA associés à l'initiation de la marche (Caderby et al. 2013). Plus spécifiquement, ces modifications permettaient aux sujets d'atteindre une vitesse de progression invariante et équivalente à celle obtenue lors de la situation de contrôle (initiation de la marche sans charge). Des résultats similaires ont par la suite été observés sur la stabilité posturale, estimée par le biais de la « marge de stabilité » (Hof et al. 2005), qui est restée inchangée malgré

une distribution asymétrique du poids du corps induite par le port d'une charge additionnelle sur un côté du corps ou de l'autre (Caderby et al. 2017). Cette invariance de la stabilité posturale, rendue possible grâce à la modulation adaptative des APA, a également été mise en évidence récemment lors d'un protocole d'initiation de la marche avec enjambement d'obstacles de taille et de hauteur changeantes (Yiou et al. 2016a, b).

Enfin, un ensemble de travaux portant sur les mouvements d'élévation du membre supérieur (Nougier et al. 1999), de flexion de la cuisse (Yiou et al. 2012b) ou encore d'initiation de la marche (Delval et al. 2005 ; Yiou et al. 2016b), a montré que la contrainte temporelle modifiait également de façon adaptative les paramètres spatio-temporels des APA. Plus spécifiquement, ces travaux ont montré de manière consistante que la durée des APA dans chacune de ces conditions était drastiquement réduite en situation de contrainte temporelle forte (les participants devaient alors exécuter le mouvement le plus rapidement possible après un signal de départ), comparativement à la situation contrôle (pas de contrainte temporelle). Cette réduction de durée était systématiquement compensée par une augmentation de l'amplitude des APA, permettant d'atteindre une performance motrice et une stabilité optimale.

### **1.1.6. Ajustements posturaux simultanés**

Bien que leur existence soit connue de longue date (Elftman 1939 ; Floyd et al. 1955 ; Lundervold 1951), ce n'est que très récemment que l'étude des ajustements posturaux simultanés a fait l'objet de recherches expérimentales systématiques. Une des raisons à cela est sans doute la difficulté de leur mise en évidence sur le plan expérimental, en comparaison aux APA et aux ajustements posturaux correcteurs. En effet, comme il a été dit à plusieurs reprises dans ce document, l'exécution du mouvement focal induit des forces internes qui sont (quasi)instantanément transmises le long de la chaîne posturale jusqu'aux surfaces d'appui. Les signaux périphériques

issus des outils classiques de la biomécanique, et en particulier des plateformes de force, ne permettent pas de distinguer directement les forces réactives induites par le mouvement de la chaîne focale de celles induites par le mouvement de la chaîne posturale. En effet, les signaux recueillis sont « globaux », i.e. résultant de la sommation des deux composantes motrices de la tâche (posturale et focale). Ce problème expérimental ne se pose pas pour l'analyse des APA et des ajustements posturaux correcteurs, puisque celles-ci sont bien découplées temporellement. Pour résoudre ce problème expérimental, un modèle biomécanique original du corps humain basé sur la méthode des ellipses de Lissajous a été développé au sein du laboratoire CIAMS ; ce modèle a permis de dissocier la dynamique induite par les composantes focales et posturales de la tâche, et donc d'investiguer les caractéristiques biomécaniques des ajustements posturaux simultanés conjointement à celles des APA et ajustements posturaux correcteurs. Ce modèle a été appliqué lors de tâches de pointage réalisées en posture assise (Fourcade et al. 2014, 2015, 2016) et lors de l'initiation d'un pas rapide (Memari et al. 2020).

Ainsi, lors du pointage avec le membre supérieur en position assise, Fourcade et al. (2014, 2016) ont montré que, comme pour les APA, les ajustements posturaux simultanés avaient une double fonction. Mais, contrairement aux APA, cette double fonction s'exerçait suivant le même axe (axe antéropostérieur, seul axe considéré dans cette étude). Selon l'instant considéré, les ajustements posturaux simultanés ont une fonction stabilisatrice, s'opposant alors aux effets perturbateurs du mouvement focal, ou une fonction d'assistance à la performance, en favorisant le développement des forces propulsives, notamment au début du mouvement volontaire (i.e. lors de l'accélération du bras vers l'avant). De plus, comme pour les APA, ces auteurs ont montré que l'amplitude des ajustements posturaux simultanés augmentait avec la vitesse du mouvement (Fourcade et al. 2016). Des résultats similaires ont été mis en évidence lors de l'exécution d'un pas rapide vers l'avant (Fourcade and Memari 2015 ; Memari et al. 2020). Sur la base de ces résultats, il a été proposé que les ajustements

posturaux simultanés sont programmés en fonction de la vitesse du mouvement volontaire et complètent les APA dans la réalisation du mouvement.

### **1.1.7. Ajustements posturaux correcteurs**

Les « ajustements posturaux correcteurs », ou consécutifs, désignent les phénomènes dynamiques posturaux ayant lieu après la fin du mouvement focal. Memari et al. (2013) ont analysé et comparé les caractéristiques des APA et des ajustements posturaux correcteurs lors de l'initiation d'un pas de longueur variable (petit et grand pas). Les résultats ont montré que la durée et l'amplitude des ajustements posturaux correcteurs sont plus importantes que celles des APA. De plus, avec l'augmentation de la longueur du pas et de la vitesse de progression, l'amplitude des ajustements posturaux correcteurs augmente, en parallèle de celle des APA. Sur la base de ces résultats, les auteurs ont proposé que ces deux types d'ajustements posturaux résultaient d'un même programme moteur.

De même, Krishnan and Aruin (2011) ont montré, au moyen de leur paradigme de perturbation posturale appliquée au niveau des épaules en station debout, que le contrôle de l'équilibre, dans ces conditions, impliquait systématiquement le développement d'APA et d'ajustements posturaux correcteurs. Ces auteurs ont également montré que lorsque les participants étaient privés d'informations visuelles, ils ne développaient quasiment plus d'APA, car ne pouvant plus prédire le déséquilibre à venir. Pour maintenir leur équilibre, les participants développaient alors davantage d'ajustements posturaux correcteurs. Les auteurs ont ainsi pu mettre en évidence une corrélation négative entre l'amplitude des APA et celle des ajustements posturaux correcteurs, i.e. plus l'amplitude des APA était faible, plus celle des ajustements posturaux correcteurs était élevée. Les effets de l'avancée en âge sur la relation entre ces deux types d'ajustements posturaux ont été étudiés par la même équipe (e.g. Kanekar and Aruin 2014). En employant un paradigme assez similaire de perturbation posturale, Kanekar and Aruin (2014) ont montré que l'amplitude des APA était



diminuée et que les délais d'activation des muscles posturaux étaient augmentés chez les personnes âgées par rapport aux jeunes adultes. De plus, malgré un recours plus important aux ajustements posturaux correcteurs, les sujets âgés étaient significativement moins stables que les jeunes adultes après une perturbation similaire. De façon intéressante, la même équipe (Aruin et al. 2015; Aruin 2016) a par la suite montré que les capacités d'anticipation posturale des personnes âgées pouvaient être améliorées grâce à un entraînement spécifique de lancers de ballons lestés (*medecine ball*). Cette amélioration leur permettait de réduire la part des ajustements posturaux correcteurs dans le contrôle postural et ainsi, de se rapprocher du comportement postural des jeunes adultes. Cette modification du comportement postural s'est traduite par une amélioration de l'équilibre. L'amélioration des APA chez les personnes âgées grâce à l'entraînement sera discuté plus en détail dans la suite de l'exposé (chapitre 1.5).

Pour résumer, les ajustements posturaux consécutifs permettraient ainsi de renforcer l'action contre-perturbatrice des APA, voire de remplacer cette action s'il n'y a pas suffisamment d'APA. De ce point de vue, les APA peuvent être perçus comme une première « ligne de défense » de l'équilibre postural, alors que les ajustements posturaux simultanés puis les ajustements posturaux correcteurs constituent la « seconde ligne de défense ». La capacité de notre système nerveux central à générer ces ajustements posturaux correspond à la « capacité posturo-cinétique ». Ce concept biomécanique de capacité posturo-cinétique est l'objet du chapitre suivant.

### **1.1.8. Concept de « Capacité posturo-cinétique »**

La capacité posturo-cinétique a été définie par Bouisset comme étant la capacité du système nerveux central à développer des phénomènes posturaux contre-perturbateurs face à une perturbation posturale donnée, et ainsi d'en atténuer ses effets négatifs sur la stabilité du corps (Bouisset and Zattara 1981, 1987 ; Bouisset and Le Bozec 2002 ; Bouisset and Do 2008). Ce concept de capacité posturo-cinétique est

originellement lié aux perturbations posturales associées au mouvement volontaire, telle que celle induite lors de l'élévation du membre supérieur en posture debout (Bouisset and Zattara 1981, 1987). Ce concept a été étendu à la compensation de perturbations endogènes « non volontaires », telles que celles induites par la respiration, les battements cardiaques et les mouvements péristaltiques. Comme nous l'avons évoqué au début de ce manuscrit, ce type de perturbations endogènes, quoique très faibles, nécessite également le développement de phénomènes dynamiques contre-perturbateurs afin que l'équilibre postural soit maintenu (e.g. Hamaoui et al. 2002). Le concept moderne de « synergie » fait d'ailleurs référence historiquement à ces mouvements contre-perturbateurs de la chaîne posturale mobilisant la tête, le bassin et les genoux, qui sont associés aux perturbations cycliques induites par les mouvements d'ampliation de la cage thoracique et le refoulement des viscères au cours de la respiration (Gurfinkel 1971).

Le concept de capacité posturo-cinétique met l'accent sur l'importance de déclencher les phénomènes posturaux *avant même* la survenue de la perturbation afin d'optimiser l'efficacité de la contre-perturbation posturale. Ce type d'adaptations correspond aux APA que nous avons décrits précédemment, et fait écho aux mécanismes d'« équilibration-action » décrits par André-Thomas (1940). Le concept de capacité posturo-cinétique met également l'accent sur le fait que tous les ajustements posturaux font partie d'un processus dynamique impliquant la mobilité de la chaîne posturale. Selon ce concept, tout facteur altérant la mobilité dynamique des segments posturaux (e.g. avec l'avancée en âge, Vandervoort et al. 1992, et, à un degré supérieur, la maladie de Parkinson, Halliday et al. 1998), aurait un impact négatif sur la stabilité posturale mais également sur la performance motrice elle-même (la vitesse maximale d'exécution ou la force maximale, par exemple). Autrement dit, la performance du mouvement focal ainsi que la stabilité posturale, dépendent de la capacité du système nerveux central à générer des ajustements posturaux, et en particulier des APA. Ce concept de capacité posturo-cinétique est en accord avec le modèle classique du

mouvement volontaire développé par Hess (1943) que nous avons décrit au chapitre 1.1.4.

De nombreux résultats expérimentaux témoignent de ce lien étroit entre mobilité dynamique de la chaîne posturale et performance motrice (e.g. Le Bozec and Bouisset 2004 ; Diakhaté et al. 2013 ; Delafontaine et al. 2015, 2017, 2018 ; Alamini-Rodrigues and Hamaoui 2017 ; Hamaoui and Alamini-Rodrigues 2017). Par exemple, des expérimentations conduites en posture assise ont montré que lorsque la mobilité du bassin est augmentée grâce à une réduction de la surface de contact au niveau des cuisses (les participants sont assis au bord de l'assise), la vitesse maximale de pointage vers une cible avec le membre supérieur augmente (Lino et al. 1992 ; Teyssède et al. 2000), de même que la force de poussée isométrique maximale d'une barre dynamométrique (Le Bozec and Bouisset 2004). Ces résultats ont été corroborés avec la tâche consistant à se lever à partir de la posture assise (*sit-to-stand*) et lors de la flexion du tronc à vitesse maximale (Diakhaté et al. 2013). Dans chacune de ces expérimentations, la plus grande mobilité de la chaîne posturale, celle du bassin en l'occurrence, était associée à une intensification de la dynamique posturale anticipatrice.

A l'inverse, il a été montré que le blocage expérimental de certaines articulations de la chaîne posturale par le biais d'une attelle rigide, responsable d'une hypomobilité articulaire, entraînait une dégradation de la capacité posturo-cinétique. Par exemple, lors de la tâche de *sit-to-stand*, Hamaoui and Alamini-Rodriguez (2017) et Alamini-Rodrigues and Hamaoui (2017) ont montré que la durée des APA était augmentée et la performance motrice réduite lorsque les sujets portaient une attèle au niveau lombaire ou cervical, comparativement à la situation contrôle sans port d'attèle. Autrement dit, la durée de la composante posturale était augmentée pour, au final, atteindre une performance motrice moindre, témoignant ainsi d'une diminution de l'efficacité des APA. A noter qu'Hamaoui et al. (2004) avaient déjà montré au préalable que la perturbation chronique de la mobilité du rachis lombaire chez les sujets lombalgiques, s'accompagnait d'une amplification des oscillations posturales lors du

maintien de la posture debout. Enfin, dans la même veine, les travaux récents de Delafontaine et al. (2015, 2017, 2018) ont montré que la restriction expérimentale de la mobilité de la cheville ou du genou de la jambe oscillante et/ou portante, s'accompagnaient d'une dégradation de l'organisation posturale et de la performance motrice lors de l'initiation de la marche.

## **1.2. Organisation neurophysiologique des APA**

### **1.2.1. Origine centrale des APA**

Les structures nerveuses à l'origine des APA ne sont pas encore bien connues. Dans un article de revue, Massion (1992) soulignait que les neurones impliqués dans le contrôle des APA pourraient être localisés au niveau de la formation réticulée (structure située dans le tronc cérébral) et de la moelle épinière. Cette hypothèse s'appuie sur des travaux réalisés chez le chat par Luccarini et al. (1990), et repris plus tard par Schepens and Drew (2003, 2006). Ces derniers ont montré que des neurones réticulo-spinaux pourraient être impliqués dans la régulation du tonus antigravitaire et le contrôle des APA. Chez l'Homme, le tronc cérébral, le cervelet, l'aire motrice primaire, l'aire pré motrice, l'aire motrice supplémentaire et les réseaux neuronaux situés entre le cortex frontal et les ganglions de la base pourraient être impliqués dans l'organisation posturale lors de l'initiation de la marche, mais peu d'études se sont intéressées aux rôles spécifiques de chacune de ces régions (Richard et al. 2017 ; Takakusaki 2017).

Richard et al. (2017) ont montré que la perturbation de l'aire motrice supplémentaire par stimulation transcrânienne entraînait une diminution de la durée des APA et une augmentation de l'activité du muscle soléaire du côté de la jambe oscillante et de celle du tibial antérieure du côté de la jambe portante. De plus, l'amplitude du déplacement latéral du centre des pressions lors des APA avait tendance à augmenter, tandis que les paramètres spatio-temporels de la phase d'exécution restaient inchangés. En employant la même technique, ces auteurs ont également

montré que la perturbation du cervelet ne modifiait pas la durée ou l'amplitude des APA mais perturbait la coordination des muscles posturaux en augmentant les co-activations musculaires entre le soléaire et le tibial antérieur, notamment du côté de la jambe portante. De plus, la durée de la phase d'exécution a significativement diminué. Ces résultats suggèrent des rôles distincts de ces deux structures nerveuses sur les APA. D'une part, l'aire motrice supplémentaire serait impliquée dans la programmation de la durée et probablement de l'amplitude des APA et, d'autre part, le cervelet dans les coordinations musculaires et le couplage des APA avec la phase d'exécution. Ces résultats corroborent partiellement ceux de Jacobs et al. (2009a), qui ont également montré une diminution de la durée des APA lors de l'initiation de la marche chez des patients parkinsoniens suite à la perturbation de l'aire motrice supplémentaire par stimulation transcrânienne, témoignant de son implication dans la synchronisation temporelle des APA. Cependant, il est à noter que Jacobs et al. (2009a) n'ont observé aucun effet sur l'amplitude des APA, contrairement à Richard et al. (2017) et Bolzoni et al. (2015). Ces derniers ont aussi montré l'implication de l'aire motrice supplémentaire dans la régulation de l'amplitude des APA lors d'un mouvement volontaire de flexion du doigt.

### **1.2.2. Organisation centrale des APA**

Bien que les structures nerveuses impliquées dans les APA ne soient pas encore bien identifiées, deux modes d'organisation centrale des APA ont été proposés dans la littérature : le mode « hiérarchique » et le mode « parallèle » (Massion 1992, 1994 ; Slijper 2002). Dans le mode hiérarchique, les APA et le mouvement focal sont les conséquences d'une seule commande motrice. Autrement dit, les composantes posturale et focale de la tâche sont exécutées de manière simultanée au niveau cérébral (Massion 1992 ; Slijper et al. 2002 ; Massion et al. 2004) et sous contrôle d'un seul programme moteur (Aruin and Latash 1995). Le modèle hiérarchique serait principalement utilisé en condition dite « réactive », i.e. lors de tâches réalisées avec un

temps de réaction (le sujet doit exécuter un mouvement en réponse à un signal sonore ; Paillard 1990). Dans ces conditions, plusieurs études ont montré que les APA apparaissent presque simultanément avec le début du mouvement focal (Nougier et al. 1999 ; Slijper 2002 ; Ilmane and LaRue 2008, 2011).

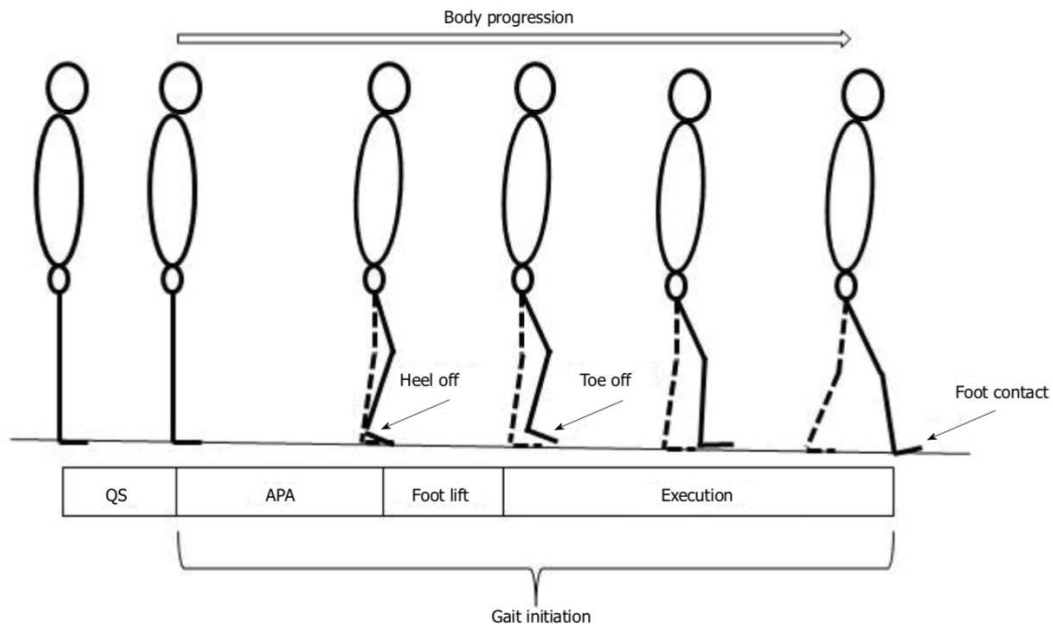
Dans le mode parallèle, les APA et le mouvement volontaire résultent de programmes moteurs distincts. Ce mode d'organisation expliquerait ainsi la présence des APA avant même le début du mouvement focal (Belen'kiï et al. 1967 ; Bouisset and Zattara 1981, 1987). De plus, ce mode de coordination entre la posture et le mouvement permettrait de moduler les APA sans affecter les paramètres du mouvement focal (Ilmane and Larue 2011 ; Slijper et al. 2002 ; Robert et al. 2007). Le mode parallèle correspondrait aux mouvements dits « auto-initiés » (i.e. réalisés sans contrainte temporelle ; Paillard 1990). En effet, pour une tâche exécutée en situation auto-initiée, les APA surviennent avant le début du mouvement focal. Le temps disponible pour la préparation du mouvement en situation auto-initiée permettrait cette dissociation en deux commandes motrices.

### **1.3. Le paradigme de l'initiation de la marche**

#### **1.3.1. Définition et enjeux**

L'initiation de la marche correspond à la phase de transition entre la position debout et une marche stable (Carlsöö 1966 ; Mann et al. 1979 ; Brenière et al. 1987 ; Jian et al. 1993 ; Yiou et al. 2017). Pour initier la marche, une jambe est décollée du sol et dirigée vers l'avant, c'est la jambe oscillante, tandis que la jambe opposée est la jambe portante. L'initiation de la marche se divise classiquement en 3 phases : une phase posturale qui précède le décolllement du talon (qui correspond aux APA), suivie d'une phase de décolllement du pied qui se termine au moment où les orteils du pied de la jambe oscillante quittent le sol (le poids du corps est transféré vers la jambe

portante à ce moment-là), et une phase d'exécution (en appui unipodal sur la jambe portante) qui se termine lorsque le pied de la jambe oscillante touche le sol (Figure 6).



**Figure 6 : Représentation des différentes phases de l'initiation de la marche.** *Body progression* : déplacement du corps, QS : position initiale ; APA : ajustements posturaux anticipateurs, *Foot lift* : phase de décollement du pied de la jambe oscillante, *Heel off* : décollement du talon, *Toe off* : décollement des orteils, *Execution* : phase d'exécution du premier pas, *foot contact* : pose du pied de la jambe oscillante au sol. D'après Yiou et al. (2017).

L'initiation de la marche est une tâche fonctionnelle classiquement utilisée dans la littérature pour étudier la façon dont le système nerveux central contrôle l'équilibre lors d'une tâche locomotrice impliquant le déplacement de l'ensemble du corps avec modification de la base de support. C'est une tâche complexe pour le système de contrôle postural car elle nécessite l'intégration de nombreuses informations provenant des systèmes proprioceptif (Ruguet et al. 2008, 2010 ; Chastan et al. 2010 ; Vinti et al. 2010 ; Lin and Yang 2011 ; Mouchnino et al. 2012 ; Mouchnino and Blouin 2013), vestibulaire (Sasaki et al. 2001 ; Henriksson et al. 2011) et visuel (Reynolds and Day 2005 ; Marigold 2008 ; Chapman et al. 2012), ainsi qu'une coordination complexe entre un ensemble de muscles répartis dans tout le corps (Bouisset and Do 2008 ; Honeine et al. 2013, 2014, 2016 ; Yiou et al. 2017). De ce point de vue, l'initiation de la

marche est une tâche locomotrice à risque de chute pour les personnes présentant des déficits sensoriels et moteurs (Robinovitch et al. 2003). Les différents mécanismes impliqués dans le contrôle de l'équilibre lors de l'initiation de la marche sont décrits dans les paragraphes suivants.

### **1.3.2. Les mécanismes de contrôle de l'équilibre lors de l'initiation de la marche**

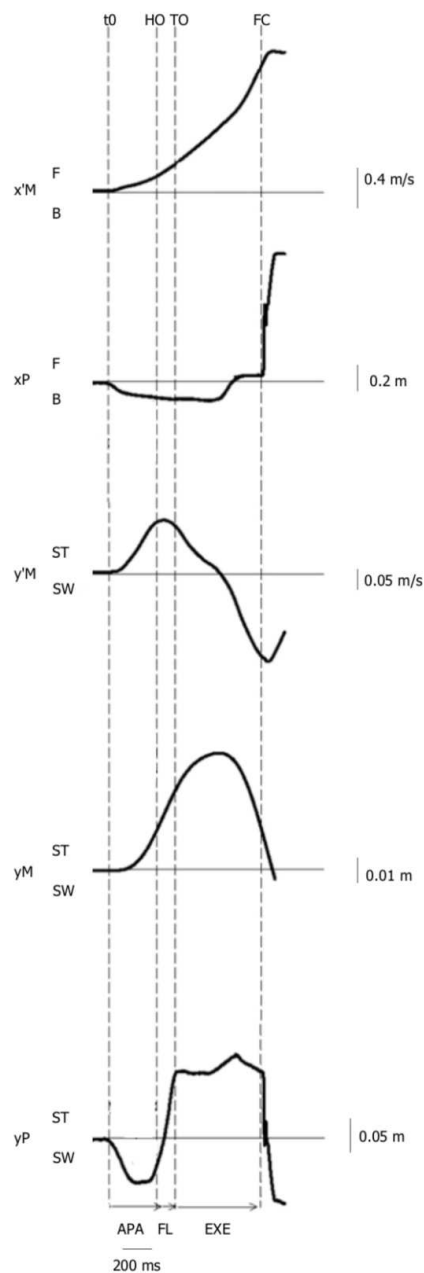
#### **1.3.2.1. Les ajustements posturaux anticipateurs (APA)**

Comme il a été dit précédemment, le fait de décoller le pied du sol pour déplacer la jambe oscillante vers l'avant induit une diminution drastique de la base posturale dans le plan frontal (Figure 5). Pour s'opposer à cette source d'instabilité potentielle, le système nerveux développe des APA qui se manifestent par un déplacement latéral du centre des pressions vers la future jambe oscillante (Figure 7). Ce déplacement permet de propulser le centre de masses dans le sens opposé, i.e. vers la jambe portante (Yiou et al. 2016b, a ; Honeine et al. 2016), et ainsi de réduire l'écart entre le centre des masses et le centre des pressions lors de la phase d'exécution. Ce déplacement médiolatéral du centre des pressions correspond à une mise en charge de la future jambe oscillante que l'on attribue à une activation des abducteurs de hanche du même côté. De plus, la hanche et le genou de la jambe portante sont légèrement fléchis lors des APA, ce qui facilite cette décharge et complète l'action des abducteurs de hanche de la jambe oscillante (Honeine et al. 2016). La flexion du genou de la jambe portante est favorisée par une inhibition bilatérale des soléaires et une plus forte activation du tibial antérieur ipsilatéral par rapport à la jambe oscillante. La flexion de la hanche est quant à elle favorisée par l'activation du droit fémoral de la future jambe de d'appui.

Sur l'axe antéropostérieur, les APA se manifestent par à un déplacement du centre des pressions vers l'arrière (Figure 7). Ce recul permet de créer un décalage entre le centre des pressions et le centre des masses, et ainsi de générer les forces



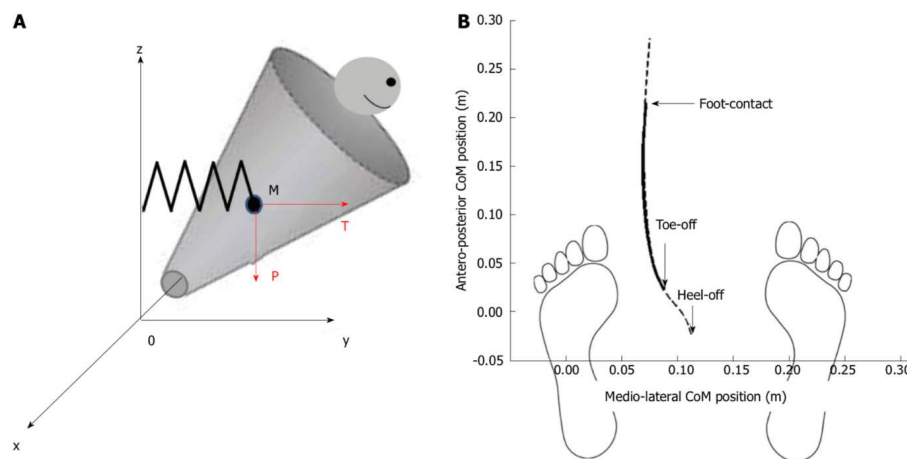
propulsives nécessaires pour atteindre la longueur de pas et de vitesse de déplacement désirées (Brenière et al. 1987 ; Lepers and Brenière 1995). Ce déplacement vers l'arrière du centre des pressions est possible grâce à une inhibition bilatérale des fléchisseurs plantaires (principalement le muscle soléaire), suivie par une activation des fléchisseurs dorsaux (principalement le tibial antérieur) (Crenna and Frigo 1991 ; Brenière and Do 1991). L'instabilité posturale au cours de l'initiation de la marche peut également être minimisée en réduisant l'amplitude des APA selon cette direction, avec pour effet une diminution de la longueur du pas.



**Figure 7 : Exemple de tracés biomécaniques chez un jeune adulte initiant la marche à vitesse maximale.** Sur l'axe antéropostérieur ( $x$ ),  $x'M$  : vitesse du centre des masses,  $xP$  : déplacement du centre des pressions, F (*forward*) : vers l'avant, B (*backward*) : vers l'arrière. Sur l'axe médiolatéral ( $y$ ),  $y'M$  : vitesse médiolatérale du centre des masses,  $yM$  : déplacement médiolatéral du centre des masses,  $yP$  : déplacement médiolatéral du centre des pressions, ST (*stance limb*) : vers la jambe portante, SW (*swing limb*) : vers la jambe oscillante. Sur les lignes pointillées,  $t_0$  : début des variations observables sur les tracés, HO (*heel off*) : décollement du talon de la jambe oscillante, FO (*foot off*) : décollement du pied de la jambe oscillante, FC (*foot contact*) : contact au sol du pied oscillant. Au niveau des flèches, APA : ajustement posturaux anticipateurs, FL (*foot lift*) : phase de décollement du pied, EXE : phase d'exécution. D'après Yiu et al. (2017).

### 1.3.2.2. La raideur de la jambe portante

Un autre mécanisme déterminant la stabilité posturale au cours de l'initiation de la marche est la régulation de la raideur de la jambe portante au cours de la phase d'exécution. Pour mettre en évidence ces liens entre raideur et stabilité posturale, un modèle mécanique original du corps, basé sur des études antérieures (MacKinnon and Winter 1993 ; Lyon and Day 1997, 2005), a été développé par Yiou et al. (2016a). Dans ce modèle, le corps humain est assimilé à un pendule inversé représenté sous forme d'un cône pivotant autour d'un point fixe (Figure 8).



**Figure 8 : modèle mécanique représentant le corps lors de la phase d'exécution de l'initiation de la marche.** A : modèle mécanique du pendule inversé. Lors de la phase unipodale (phase d'exécution), le corps est représenté comme un cône inversé pivotant autour d'un point fixe  $O$ . Il peut se déplacer selon 5 degrés de liberté (trois translations, deux rotations). Le centre des masses  $M$  chute latéralement sous l'influence de la force de gravité  $P$  et de la force de restitution  $T$ . B : Tracés biomécaniques du centre des masses. La ligne pointillée représente les données expérimentales recueillies lorsqu'un sujet initie la marche sur une plateforme de force. La ligne pleine représente les données théoriques obtenues à partir du modèle mécanique. Le modèle montre une excellente reproductibilité des données expérimentales. D'après Yiou et al. (2016a, 2017)

Dans ce modèle, le centre des masses tombe latéralement sous l'effet de deux forces qui s'opposent lors de la phase d'exécution : d'un côté, la force de gravité ( $P$ ), avec  $P = mg$  (où  $m$  est la masse du solide et  $g$  l'accélération de la gravité) et de l'autre, une force de restitution  $T$  correspondant à l'activité des muscles de la jambe portante,

avec  $T=k|y_M|$ , où  $k$  représente la raideur des muscles de la jambe portante lors de la phase d'exécution et  $|y_M|$  est la valeur absolue de  $y_M$ , soit le déplacement du centre des masses sur l'axe médiolatéral, qui est systématiquement orienté vers la jambe oscillante lors de la phase d'exécution. La position et la vitesse initiale du cône correspondent à la position et à la vitesse du centre des masses du sujet au moment du décolllement du pied (juste avant la phase unipodale). L'ajout de la force de restitution est nécessaire pour contrôler la vitesse initiale du centre des masses au décolllement du pied. En effet, sans l'ajout de cette force, le cône chuterait systématiquement du côté de la jambe portante sous l'effet des APA médiolatéraux. La figure 8 montre que les tracés biomécaniques obtenus expérimentalement avec des sujets initiant la marche sur une plateforme de force se superposent à ceux obtenus avec le modèle mécanique. La meilleure correspondance entre les données expérimentales (ligne pointillée) et les données théoriques (ligne pleine) est obtenue pour une raideur  $k$  de la jambe portante autour de  $1000 \text{ N/m}$ . Ce qui correspond à une force de restitution d'environ  $T= 50\text{N}$  appliquée au centre des masses du participant.

Grâce à ce modèle, il a été montré que la raideur de la jambe portante lors de la phase d'exécution pouvait influencer l'amplitude de la chute médiolatérale du centre des masses et donc la stabilité posturale (Yiou et al. 2016a). Cette raideur globale de la jambe portante peut théoriquement être ajustée par l'intermédiaire des co-activations musculaires entre les muscles agonistes et antagonistes qui mobilisent et stabilisent les articulations de la hanche, du genou ou de la cheville. De ce point de vue, un contrôle optimal de la raideur et donc du niveau de co-activation musculaire au niveau des articulations de la jambe portante permettrait de minimiser la chute latérale du centre des masses, et ainsi d'accompagner le développement des APA médiolatéraux. En ce sens, le contrôle de la raideur au cours de l'exécution du pas peut être assimilé à une forme d'ajustement postural simultané.

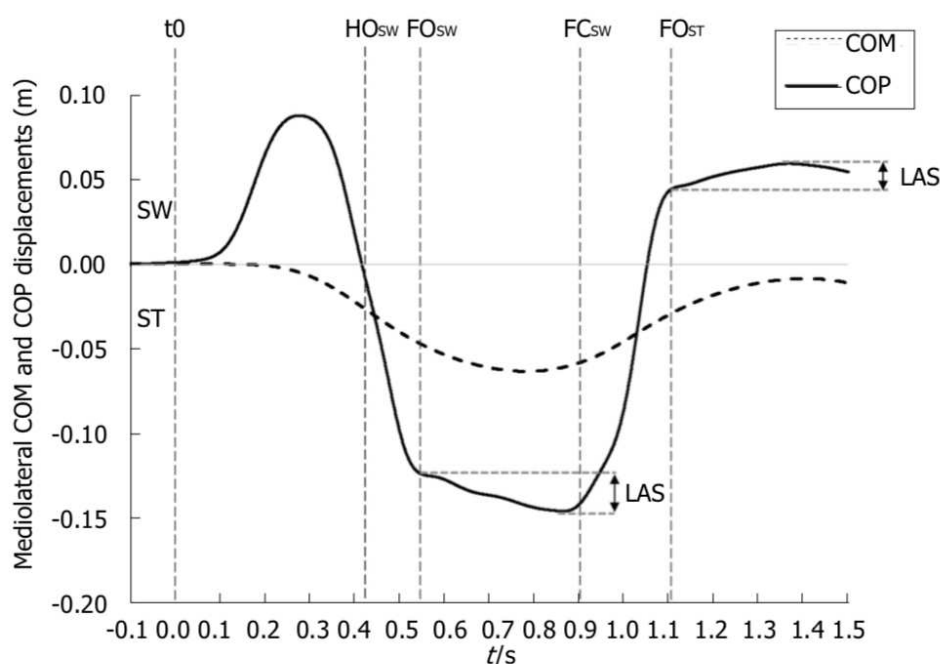
### 1.3.2.3. Positionnement du pied et stratégie latérale de cheville

Bien que la modulation des APA sur l'axe médiolatéral et de la raideur de la jambe portante influencent l'amplitude de la chute médiolatérale du centre des masses vers la jambe oscillante lors de la phase d'exécution, ces mécanismes ne sont pas toujours suffisants pour obtenir une stabilité optimale lors de l'initiation de la marche (Jian et al. 1993 ; Lyon and Day 1997 ; McIlroy and Maki 1999 ; Yiou et al. 2016a). Dans ce cas, la stabilité médiolatérale doit être réajustée afin de poursuivre la progression de façon sécuritaire.

Le premier mécanisme de récupération de l'équilibre lors de la phase unipodale est le placement du pied oscillant au sol de façon à ajuster la taille de la base de support (Townsend 1985 ; MacKinnon and Winter 1993). Cette action permet d'élargir la base de support au moment de la pose du pied oscillant au sol et de déplacer le centre des pressions latéralement, au-delà du centre des masses. Ainsi, la chute latérale du corps peut être ralentie et le centre des masses accéléré dans le sens opposé, vers la jambe portante (Figure 9).

Il a été montré que le placement du pied est activement régulé par le système nerveux central pour restaurer et contrôler l'équilibre dans le plan frontal (Bauby and Kuo 2000 ; Donelan et al. 2004 ; O'Connor and Kuo 2009 ; Collins and Kuo 2013). Le placement du pied est principalement ajusté par l'intermédiaire des abducteurs de hanche de la jambe oscillante en fonction de la position et de la vitesse du centre des masses (Rankin et al. 2014). Par exemple, lorsque les sujets initient la marche rapidement, le centre des masses a tendance à chuter davantage du côté de la jambe oscillante (Caderby et al. 2014). Néanmoins, chez des sujets en bonne santé, il est possible de compenser cette instabilité en augmentant la largeur du pas (et donc la base de support) de façon à ce que l'équilibre dynamique sur l'axe médiolatéral reste inchangé. Cette ajustement fin du placement du pied peut être altéré chez des patients souffrant de troubles sensoriels et/ou moteurs, par exemple chez des sujets amputés au-dessus du genou (Hof et al. 2007).

Un deuxième mécanisme peut intervenir après la pose du pied au sol (phase de double appui), dans le but de corriger d'éventuelles erreurs de placement du pied. Ce mécanisme est appelé « stratégie latérale de cheville ». Une fois le pied de la jambe oscillante au sol, il est encore possible de contrôler la position médiolatérale du centre des pressions sous le pied oscillant, en activant les muscles éverseurs et inverseurs du pied (King and Zatsiorsky 2002). Cette stratégie intervient aussi, et majoritairement, au niveau du pied de la jambe portante lors de la phase unipodale (Figure 9), cependant, la marge de correction de cette stratégie est faible car limitée par la largeur du pied. A nouveau, des troubles sensoriels ou moteurs peuvent altérer ce mécanisme (les altérations des mécanismes du contrôle postural seront détaillées dans le chapitre 1.4).



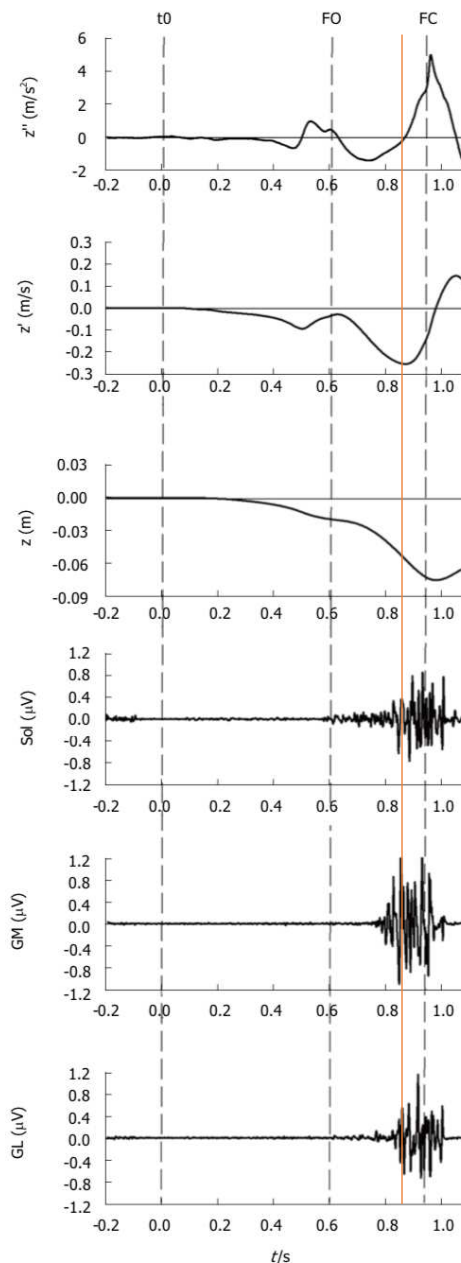
**Figure 9 : Tracés biomécaniques typiques sur l'axe médiolatéral lors de l'initiation de la marche.** Ces tracés correspondent aux déplacements médiolatéraux du centre des masses (COM) et du centre des pressions (COP) chez un sujet initiant la marche à vitesse spontanée. SW : vers la jambe oscillante, ST : vers la jambe portante, t0 : début des variations des tracés, HO (*heel off*) : décollage du talon, FO (*foot off*) : décollage du pied, FC (*foot contact*) : contact du pied au sol, LAS (*lateral ankle strategy*) : stratégie latérale de cheville. Le niveau de correction par cette stratégie correspond à la différence entre la position latérale initiale du COP et sa position latérale maximale lors de la phase unipodale. Cette différence correspond à la distance qui sépare les lignes fines pointillées (entre les flèches). D'après Yiou et al. (2017).

Pour résumer, le placement latéral du pied oscillant et la stratégie latérale de cheville sont deux mécanismes indépendants, mais qui sont couplés et coordonnés (Reimann et al. 2017). Ces mécanismes complètent les APA médiolatéraux et la raideur de la jambe portante pour le contrôle de l'équilibre du corps dans le plan frontal.

#### **1.3.2.4. Freinage vertical du centre des masses**

Chez les adultes en bonne santé, la vitesse verticale maximale du centre des masses (en valeur absolue) est atteinte avant le contact du pied au sol, environ à la moitié de la phase unipodale (phase d'exécution), puis elle est ralentie (Figure 10). Cette décélération de la chute verticale du centre des masses est liée à une augmentation de l'activité du triceps sural de la jambe portante lors de la deuxième moitié de la phase unipodale (Honeine et al. 2013).

En contrecarrant les effets de la gravité, le triceps sural joue un rôle majeur dans la régulation de la chute verticale du centre des masses, la durée de la phase d'exécution et la longueur du pas (Honeine et al. 2014). D'une part, le freinage du centre des masses permet de limiter l'amplitude des forces de réaction du sol lors de l'impact du pied oscillant au sol (Welter et al. 2007). D'autre part, les forces produites par le triceps sural lors de la deuxième moitié de la phase unipodale permettent de réduire le coût énergétique de la marche, en facilitant le redressement du centre des masses lors de la phase de double appui suivante (Kuo 2007).



**Figure 10 : Freinage vertical du centre des masses lors de l'initiation de la marche.**

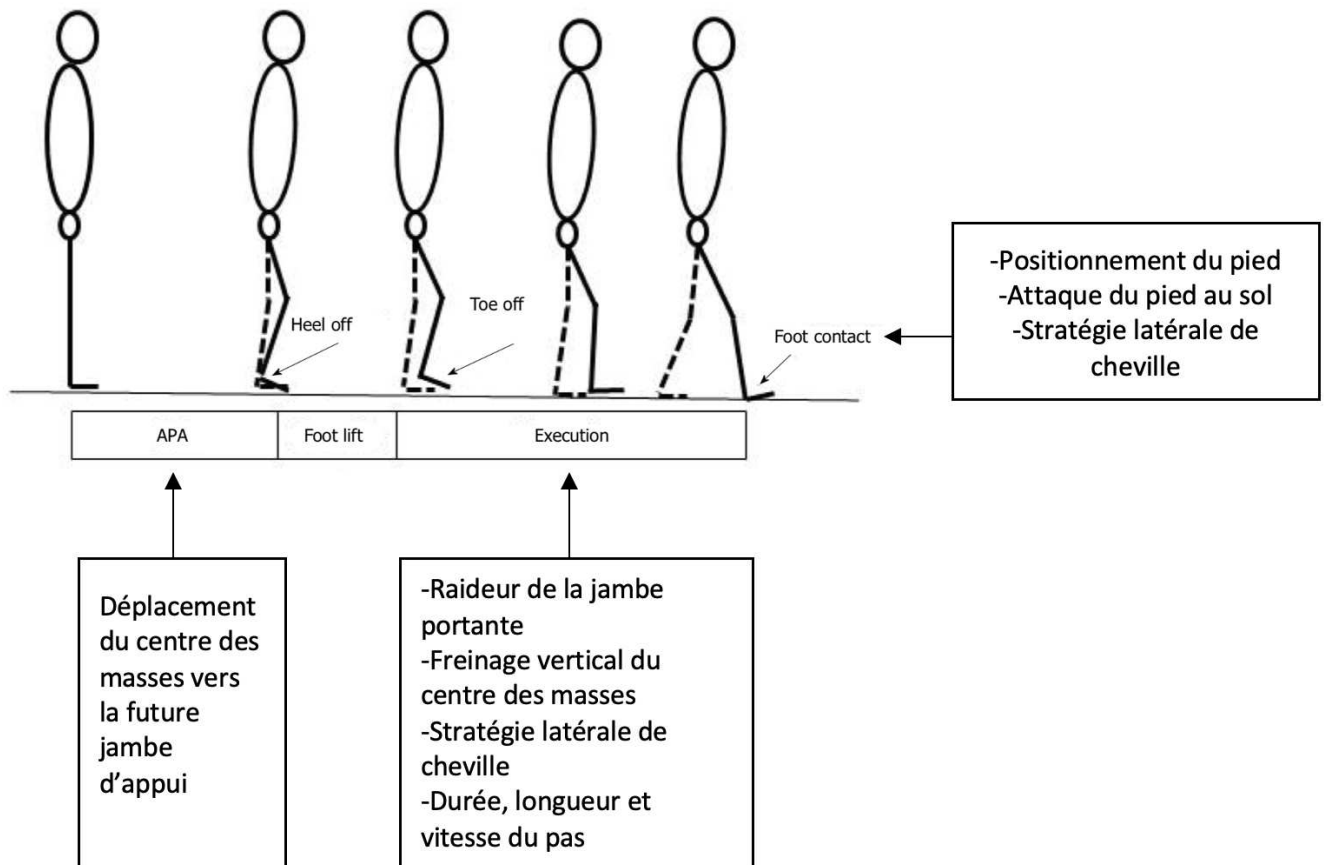
Les trois premiers graphiques montrent les tracés biomécaniques avec, l'accélération verticale du centre des masses ( $z''$ ), la vitesse verticale du centre des masses ( $z'$ ) et la position verticale du centre des masses ( $z$ ). Les trois graphiques suivants montrent l'activité électromyographique du triceps sural de la jambe portante au niveau, des muscles soléaire (Sol), gastrocnémien médial (GM) et gastrocnémien latéral (GL). Les lignes pointillées indiquent le début de l'initiation de la marche ( $t_0$ ), le décollage du pied (FO), et le contact du pied oscillant au sol (FC). L'exécution du pas s'accompagne d'une accélération du centre des masses vers le bas. Vers la moitié de la phase d'exécution, les trois muscles composant le triceps sural s'activent, ce qui coïncide avec le début du freinage vertical du centre des masses (ligne rouge verticale). D'après Yiou et al. (2017) et Honeine et al. (2013).



### **1.3.2.5. Stratégies d'« attaque » du pied au sol**

Il est connu que lors de la locomotion (marche ou course à pieds), le contact du pied oscillant au sol peut se faire de trois façons : par l'arrière du pied (le talon attaque le sol en premier), avec le pied à plat (l'avant et l'arrière du pied touchent le sol en même temps), et par l'avant du pied (les orteils touchent le sol avant le talon) (Lieberman et al. 2010 ; Daoud et al. 2012). Le fait d'attaquer le sol avec l'avant du pied permet de réduire les forces de collisions avec le sol. Ceci est rendu possible grâce à une flexion plantaire plus importante et une meilleure compliance du pied lors du contact avec le sol. La masse de corps impactant le sol au même instant est alors moins importante. Cependant, dans la plupart des études s'intéressant à l'initiation de la marche, les sujets qui se déplacent souvent pieds nus sur une plateforme de force rigide, adoptent spontanément une stratégie d'attaque par le talon quelle que soit la vitesse de progression. La stratégie d'attaque par l'avant du pied est davantage choisie lorsque la longueur de pas imposée augmente, avec un obstacle à franchir par exemple (Yiou et al. 2016a), ou lors de la descente d'une marche (van Dieën et al. 2008). Comme pour le freinage actif du centre des masses, la pose du pied joue donc un rôle d'amortissement des chocs, atténuant ainsi les effets potentiellement déstabilisants des forces de réaction du sol au moment du contact du pied oscillant au sol.

Les différents mécanismes du contrôle postural lors de l'initiation de la marche sont résumés dans la figure 11.



**Figure 11 : résumé des mécanismes de contrôle postural lors de l'initiation de la marche.** Modifié, d'après Yiou et al. (2017).

### 1.3.3. Évaluation de l'équilibre dynamique lors de l'initiation de la marche

Comme vu précédemment (chapitre 1.1.1), pour que l'équilibre soit maintenu en position debout statique, il faut que le centre des masses soit contenu dans la base de support. Cette condition de stabilité est maintenue tant qu'il y a très peu d'accélération du centre des masses. Cependant, lors de tâches locomotrices comme l'initiation de la marche, la vitesse de déplacement du centre des masses ne peut être négligée et elle est prise en considération pour évaluer la stabilité posturale (Pai and Patton 1997 ; Hof et al. 2005). Ainsi, même lorsque le centre des masses est au-dessus de la base de support, l'équilibre postural est en danger s'il est accéléré en dehors. A l'inverse, si le centre des masses est en dehors de la base de support, il est toujours

possible de récupérer l'équilibre en accélérant le centre des masses vers la base de support (Hof et al. 2005). Par exemple, lors de l'initiation de la marche, le centre des masses est accéléré vers l'avant, en dehors de la base de support, avant d'être repositionné dans la nouvelle base de support au moment de la pose du pied oscillant (Yiou et al. 2017).

Jusqu'à récemment, la stabilité lors de l'initiation de la marche était mesurée en fonction du déplacement médiolatéral du centre des masses, de sa vitesse et de la largeur du pas au moment du contact du pied oscillant au sol. Plus les valeurs de la cinématique du centre des masses étaient faibles et plus la largeur du pas était importante, plus la situation était considérée comme étant stable. Cependant, la prise en compte de ces paramètres séparément rend parfois les interprétations difficiles. Par exemple, lorsque la base de support augmente en même temps que le déplacement ou la vitesse médiolatéraux du centre des masses, il devient impossible de déterminer si la stabilité est altérée ou non.

Hof et ses collègues (Hof et al. 2005) ont donc proposé un outil d'évaluation de l'équilibre dynamique nommé « marge de stabilité » regroupant les paramètres cités précédemment. Il se définit par la distance minimale entre la limite médiolatérale de la base de support et la position médiolatérale du « centre des masses extrapolé » lors du contact du pied oscillant au sol. « Le centre des masses extrapolé » correspond à une extrapolation de la trajectoire du centre des masses à partir de son déplacement et de sa vitesse. La marge de stabilité prend ainsi en considération à la fois la position du centre des masses, sa vitesse et la base de support. L'équilibre dynamique est acquis si le centre des masses extrapolé est positionné dans la base de support, ce qui correspond à une valeur de marge de stabilité positive. Une valeur négative de la marge de stabilité signifie une instabilité latérale, ce qui nécessite un pas de côté supplémentaire pour rattraper le déséquilibre.

### **1.3.4. Conclusion sur le paradigme de l'initiation de la marche**

La stabilité de l'équilibre est mise à l'épreuve lors de l'initiation de la marche. Le système nerveux central est donc amené à développer un ensemble de mécanismes pour assurer une progression sécuritaire du corps. En accord avec le concept de capacité posturo-cinétique, chacun de ces mécanismes implique la mobilité de la chaîne posturale et en particulier celle de la cheville. Par conséquent, l'altération de cette mobilité, e.g. avec l'avancée en âge aussi bien physiologique que pathologique (en particulier avec la maladie de Parkinson), peut potentiellement être responsable d'une instabilité et de la survenue d'une chute au cours de l'initiation de la marche.

## **1.4. Effets de l'avancée en âge et de la maladie de Parkinson sur les mécanismes du contrôle postural lors de l'initiation de la marche**

Il est admis que le vieillissement physiologique est associé à une dégradation progressive du contrôle postural et de la marche (Muyinat et al. 2019). Ces changements liés à l'âge sont multifactoriels. On constate notamment une altération structurelle et fonctionnelle du système proprioceptif (Skinner et al. 1984 ; Madhavan and Shields 2005 ; Kararizou et al. 2005 , Shaffer and Harrison 2007), une baisse de la sensibilité cutanée au niveau de la voûte plantaire (Liu et al. 2005 ; Peters et al. 2016 ; Korchi et al. 2019), une diminution de l'acuité visuelle (Muyinat et al. 2019) et une dégradation du système vestibulaire (Woollacott and Tang 1997). De plus, la capacité d'intégration des signaux multi sensoriels est diminuée au niveau du cerveau, altérant la sélection adéquate des informations (Woollacott et al. 1986). En fin, il est à noter que la diminution de la vitesse de conduction nerveuse retarde les temps de conduction des afférences visuelles (Woollacott and Tang 1997) et des réflexes posturaux (Dorfman and Bosley 1979). Ces changements peuvent altérer l'équilibre et augmentent les

amplitudes des oscillations posturales (Horak et al. 1989 ; Alexander et al. 1992), davantage en cas de privation de la vue car la visio-dépendance augmente avec l'âge (Muyinat et al. 2019).

Pour les tâches locomotrices, la réduction de la vitesse de marche est le principal changement observé lors du vieillissement et un indicateur de l'autonomie fonctionnelle des patients (Perera et al. 2016 ; Cruz-Jimenez 2017). D'un point de vue biomécanique, on observe chez les personnes avancées en âge une diminution des amplitudes articulaires lors de la marche autour de la cheville et de la hanche, ainsi qu'une baisse des forces de propulsion autour de la cheville accompagnée d'une diminution des forces de réaction du sol (Boyer et al. 2017). Ces changements liés à l'âge jouent un rôle important dans les limitations fonctionnelles des personnes âgées et peuvent impacter le contrôle postural lors de l'initiation de la marche, comme dit précédemment.

La maladie de Parkinson est une maladie neurodégénérative liée à l'âge qui se manifeste principalement autour de 65-70 ans (Tysnes and Storstein 2017). Les symptômes typiques sont le ralentissement moteur, la difficulté à déclencher des mouvements volontaires, une augmentation du tonus musculaire responsable de la raideur des membres, des tremblements de repos et une instabilité posturale. De nombreux patients souffrent aussi de troubles cognitifs associés (Lee and Gilbert 2016; Tysnes and Storstein 2017). La maladie de Parkinson est une affection neurodégénérative principalement caractérisée par une perte progressive des neurones dopaminergiques au niveau de la substance noire (Chaumette et al. 2009, Corrêa et al. 2019). L'objectif principal des traitements est donc de promouvoir l'action de la dopamine déficiente par l'administration de L-DOPA, un précurseur de la dopamine. La L-DOPA favorise la synthèse de dopamine et améliore l'état des patients (Burleigh-Jacobs et al. 1997 ; Schaafsma et al. 2003 ; Bryant et al. 2011 ; Caliandro et al. 2011). Certains patients sont aussi traités par stimulations intracérébrales et d'autres traitements sont en cours de développement (Bear et al. 2015).

Il est actuellement bien reconnu que le processus d'initiation de la marche est d'un intérêt particulier pour l'étude des effets du vieillissement et de la maladie de Parkinson sur le contrôle postural (e.g. Delval et al. 2014). Aussi, les effets du vieillissement physiologique et de la maladie de Parkinson sur les mécanismes de contrôle de l'équilibre au cours de l'initiation de la marche (décrits aux paragraphes précédents) sont abordés aux chapitres suivants.

#### **1.4.1. Effets du vieillissement et de la maladie de Parkinson sur les APA**

Nous avons vu que les déplacements du centre des pressions et du centre des masses lors des APA dépendent de synergies musculaires impliquant une part étendue de la musculature striée squelettique. Ces synergies anticipatrices, dynamiques et musculaires, sont altérées chez les personnes âgées saines, à risque de chute et atteintes de la maladie de Parkinson.

Une étude récente a montré que les paramètres spatio-temporaux des APA lors de l'initiation de la marche en terrain plat et à vitesse auto initiée ne varient pas significativement avec l'avancée en âge (Lu et al. 2017). Ce résultat contraste avec l'étude de Khanmohammadi et al. (2015) qui a montré une diminution de l'amplitude et de la vitesse de déplacement du centre des pressions lors des APA sur l'axe antéropostérieur chez des personnes âgées en bonne santé, ainsi qu'un délai plus important d'activation des muscles tibial antérieur et moyen fessier. Cette différence pourrait en partie s'expliquer par les consignes données aux sujets. En effet, dans l'étude de Khanmohammadi et al. (2015), les sujets devaient initier la marche le plus rapidement possible après un signal auditif, alors qu'aucune pression temporelle n'a été appliquée dans l'étude de Lu et al. (2017). Or, nous avons vu que la contrainte temporelle impactait les paramètres spatio-temporels des APA lors de l'initiation de la marche (chapitre 1.1.5, Delval et al. 2005 ; Yiou et al. 2016b). Pour rappel, chez les sujets jeunes, la durée des APA était réduite mais compensée par une augmentation

d'amplitude des APA, afin de maintenir une performance motrice optimale. Chez les personnes âgées en bonne santé, cette compensation n'apparaît pas, probablement du fait de l'altération de la réponse musculaire (Khanmohammadi et al. 2015).

En comparant des sujets âgés sains et des sujets présentant un risque de chute important, Uemura et al. (2011) ont montré qu'il n'y avait pas de différences significatives entre ces deux populations lors de la phase posturale de l'initiation de la marche en terrain plat. En revanche, la durée des APA était plus longue chez les sujets chuteurs lors de l'initiation de la marche avec obstacle (Uemura et al. 2011). Dans la même veine, ces auteurs ont montré une augmentation de la durée des APA lors de l'initiation de la marche réalisée en situation de double tâche (i.e. compter à rebours en attendant le signal de départ) chez des patients âgés ayant peur de chuter par rapport à des personnes autonomes du même âge. Cependant, aucune différence significative entre les deux groupes lors de l'initiation de la marche en situation de simple tâche n'a été observée (Uemura et al. 2012a). Il est à noter que même chez des sujets jeunes en bonne santé, la peur de tomber peut impacter la capacité à générer des APA lors de tâche motrices (Adkin et al. 2002 ; Yiou et al. 2011a).

Chez les patients âgées présentant des déficits de hauteur et de longueur de pas, Patchay et al. (1997, 2002) ont aussi montré une diminution des forces verticales, ce qui expliquerait en partie les difficultés à décoller le pied oscillant. Ces auteurs ont proposé que ces perturbations sont susceptibles de refléter une altération des synergies musculaires au niveau des membres inférieurs. Cette hypothèse a été corroborée par les travaux de Wang et al. (2013, 2015) lors de l'initiation du pas sous pression temporelle et avec obstacle chez des personnes âgées en bonne santé. L'altération de ces synergies musculaires anticipatrices pourrait être aggravée chez patients atteints de pathologies neurodégénératives.

Une altération des APA spécifiques à la maladie de Parkinson a été mise en évidence dans la littérature de façon consistante. Ainsi, au moment de faire le premier pas, les patients parkinsoniens hésitent parfois, voire se bloquent. Ce phénomène est

plus fréquent lorsque la maladie est à un stade avancé. Ce symptôme de « freezing » se traduit par une incapacité du patient à décoller le pied du sol et s'accompagne de tremblements des genoux (Schaafsma et al. 2003 ; Hausdorff et al. 2003). Il a été montré que ces tremblements se caractérisent par de « multiples » APA qui sont observables par des mises en charge et décharges successives des jambes droite et gauche (Jacobs et al. 2009b). Ces cycles de mises en charge et décharges des jambes s'accompagnent d'activations et désactivations alternées des muscles fléchisseurs dorsaux de cheville (le tibial antérieur par exemple) et des abducteurs de hanches (le tenseur du fascia lata par exemple). Les tremblements des genoux rendent inefficaces le déplacement du centre des masses vers la jambe portante, ce qui complique le couplage des APA au déclenchement du pas. De plus, les hésitations lors de l'initiation de la marche chez les patients parkinsoniens qui sont sujets au « freezing » s'accompagnent d'une plus grande variabilité de la jambe oscillante, contrairement à des sujets en bonne santé qui ont tendance à toujours initier la marche avec la même jambe (Okada et al. 2011a). Les dysfonctionnements lors de la phase posturale de l'initiation de la marche sont généralement améliorés par les traitements à base de Levodopa (Rocchi et al. 2012). Cependant, malgré le traitement, Okada et al. (2011a, b) ont montré que le déplacement du centre des pressions sur l'axe antéropostérieur lors des APA pouvait demeurer significativement plus faible par rapport à des sujets jeunes en bonne santé. Ceci entraîne une diminution de la longueur du pas et de la vitesse de marche, alors que la durée des APA augmente. Dans ce cas, la phase posturale des patients parkinsoniens est moins efficace, ce qui traduit une dégradation de la capacité posturo-cinétique. Ce ralentissement moteur lors de l'initiation de la marche est connu depuis longtemps maintenant. Il s'aggrave avec l'avancée en âge et davantage avec la progression de la maladie de Parkinson (Halliday et al. 1998). Chez les patients qui ne sont pas sous traitement, d'autres auteurs ont montré que les déplacements médiolatéraux du centre des pressions pouvaient également être altérés (Mancini et al. 2009 ; Bonora et al. 2017).



### **1.4.2. Effets du vieillissement et de la maladie de Parkinson sur la raideur de la jambe portante**

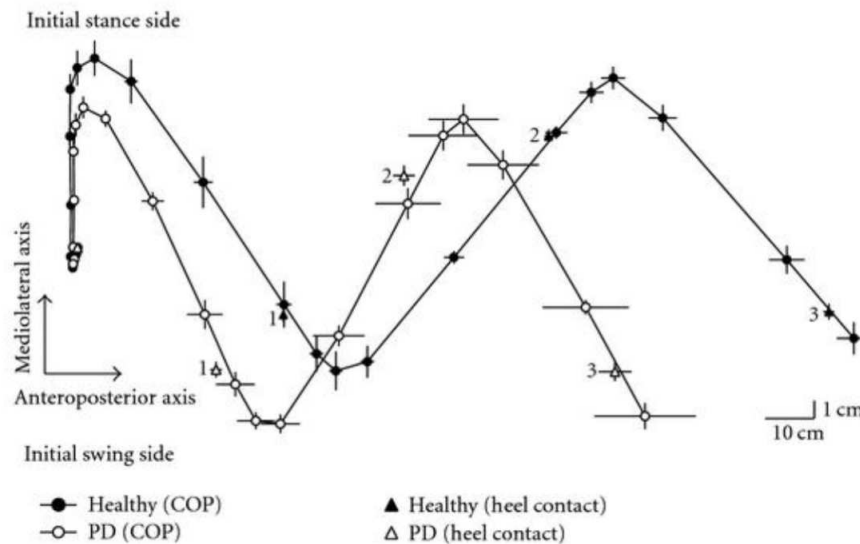
Nous avons vu au travers du modèle mécanique développé par Yiou et al. (2016a) que la raideur de la jambe portante a une influence sur la vitesse de la chute médiolatérale du centre des masses lors de la phase d'exécution, et donc sur la stabilité posturale (paragraphe 1.3.3). Or, une augmentation de la raideur au niveau des membres inférieurs est couramment observée chez les personnes âgées (Vandervoort et al. 1992), et celle-ci est exacerbée chez les patients atteints de la maladie de Parkinson (augmentation de la résistance lors de la mobilisation passive des segments corporels, Raza et al. 2019). Selon le modèle mécanique décrit précédemment, une partie des troubles de l'équilibre au cours de l'initiation de la marche chez ces personnes pourrait ainsi être attribuée à une raideur accrue des membres inférieurs. Certains auteurs ont reproduit cette raideur chez des sujets sains avec des attelles plus ou moins rigides au niveau des membres inférieurs pour étudier les effets de l'hypomobilité sur le contrôle postural chez les jeunes adultes. Par exemple, il a été montré que le port d'une attelle rigide au niveau de la cheville de la jambe portante entraîne une augmentation de la chute latérale du centre des masses (Delafontaine et al. 2017). Cependant, ce résultat ne se retrouve pas lorsque l'on entrave la mobilité du genou avec une attelle (Delafontaine et al. 2018). Dans ce cas, seule la performance motrice (mesurée par la longueur de pas et la vitesse de marche) a été altérée, mais la stabilité posturale a été préservée. Ces résultats suggèrent que la sensibilité du système postural à l'hypomobilité du membre inférieur n'est pas identique selon l'articulation concernée. La perte de mobilité autour de la cheville semble affecter davantage le contrôle postural dans le plan frontal.

### **1.4.3. Effets du vieillissement et de la maladie de Parkinson sur le placement latéral du pied et la stratégie latérale de cheville**

Comme vu précédemment (paragraphe 1.3.3), le placement latéral du pied et la stratégie latérale de cheville sont deux mécanismes coordonnés qui complètent les APA et le contrôle de la raideur de la jambe portante dans le contrôle postural sur l'axe médiolatéral. Les jeunes adultes en bonne santé sont capables d'ajuster finement la largeur du pas, de façon à maintenir une stabilité latérale inchangée même lorsque celle-ci est potentiellement modifiée, comme c'est le cas par exemple avec l'augmentation de vitesse de marche (Caderby et al. 2014). La régulation du placement latéral du pied est sous contrôle du système nerveux central et s'effectue grâce au recrutement des muscles abducteurs de hanche de la jambe oscillante, en réponse à la position et à la vitesse du centre des masses sur l'axe médiolatéral (Hof et al. 2007 ; Rankin et al. 2014). Ceci implique que ce mécanisme du contrôle postural peut être altéré chez les personnes présentant des déficits sensoriels ou moteurs. Par exemple, chez les patients amputés au-dessus du genou ou les patients ayant subi un accident vasculaire cérébral (AVC), le contrôle du placement latéral du pied est dégradé, ce qui entraîne une plus grande instabilité dans le plan frontal (Hof et al. 2007 ; Dean and Kautz 2015). De même, les personnes âgées présentent une plus grande variabilité de largeur de pas lors de l'initiation de la marche (Muir et al. 2014) et sont moins aptes que des jeunes adultes à déclencher des pas sur le côté pour rattraper les déséquilibres latéraux (Maki et al. 2000 ; Maki and McIlroy 2006). En cas de déséquilibre médiolatéral, les personnes âgées ont tendance à produire plusieurs petits pas, ou avoir recours aux membres supérieurs pour se rattraper. Cependant ces mécanismes restent moins efficaces par rapport aux jeunes adultes qui contrôlent mieux la largeur du pas en fonction du déséquilibre.

Chez les patients parkinsoniens, le premier contact du pied oscillant au sol a tendance à être dévié du côté de la jambe oscillante et vers l'arrière (Okada et al.

2011b). Autrement dit, le premier pas est plus large et plus petit que des sujets en bonne santé (Figure 12).



**Figure 12 : Déplacement moyen du centre des pressions et contact du pied au sol lors de l'initiation de la marche chez un patient parkinsonien et un sujet contrôlé.**

La ligne avec les ronds noirs correspond au trajet du centre des pressions (COP) du sujet contrôlé (*healthy* (COP)), la ligne avec les ronds blancs correspond au trajet du centre des pressions du sujet parkinsonien (PD(COP)). Les triangles correspondent aux premier (1), deuxième (2) et troisième (3) contacts du pied avec le sol (*healthy* (*heel contact*) et PD (*heel contact*)). Le placement du premier pied oscillant (1) est significativement dévié du côté de la jambe oscillante sur l'axe médiolatéral. D'après Okada et al. (2011b)

Le vieillissement est également à l'origine de modifications des mécanismes d'équilibration lors de la phase unipodale et de la pose du pied au sol. Il a été montré que la stratégie de cheville est moins sollicitée avec l'avancée en âge et chez des patients souffrant de pathologies neurologiques (Gatev et al. 1999 ; Coquisart 2012). Certains auteurs ont proposé que cette adaptation serait due à une diminution de la force des muscles autour de la cheville (Gatev et al. 1999 ; Onambele et al. 2006). Coquisart (2012) quant à lui, a proposé que les mécanorécepteurs périphériques du système somatique sensoriel impliqués dans le réflexe myotatique et les vitesses élevées de conduction de l'information nerveuse permettent aux sujets jeunes de s'organiser autour d'une stratégie latérale de cheville en sollicitant les éverseurs et

inverseurs de la cheville. A l'inverse, les sujets âgés sont quant à eux contraints d'adopter une stratégie de hanche pour compenser l'augmentation des latences de l'information nerveuse (Dorfman and Bosley 1979). Le contrôle de l'équilibre autour d'un pivot proximal permet ainsi aux personnes âgées de réduire la distance entre l'origine des informations et les centres chargés de les traiter (Coquisart 2012).

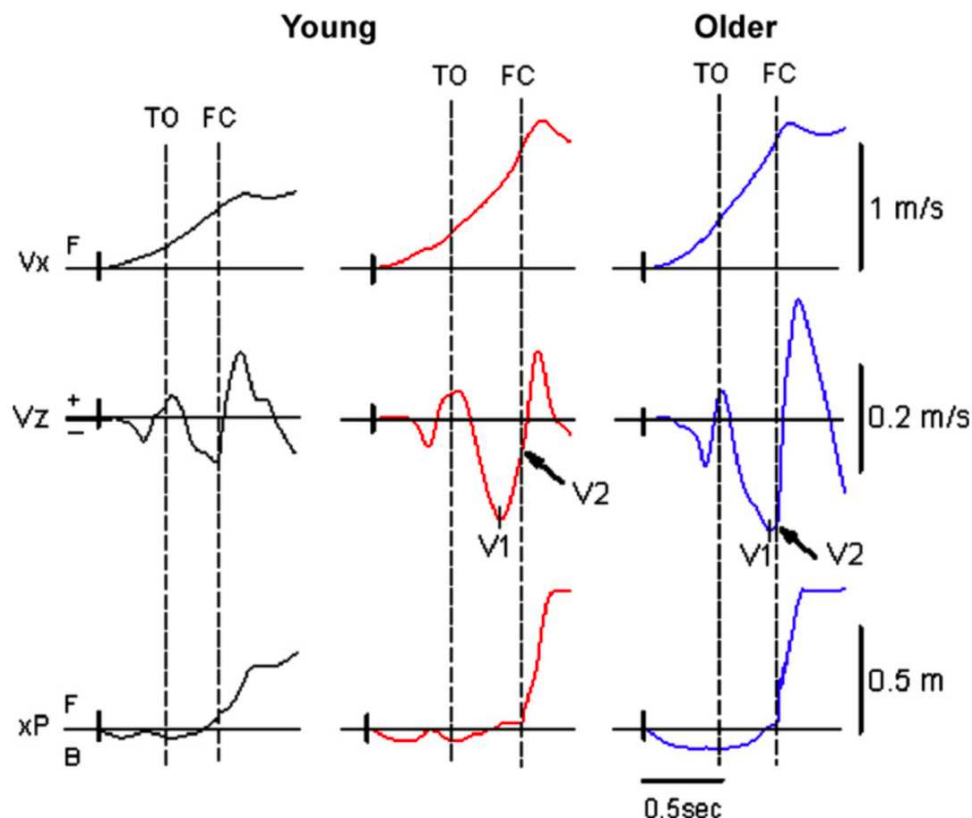
#### **1.4.4. Effets du vieillissement et de la maladie de Parkinson sur le freinage vertical du centre des masses**

Comme nous l'avons vu précédemment (Chapitre 1.3.3), la chute du centre des masses lors de la phase d'exécution de l'initiation de la marche est activement freinée sous l'action du triceps sural de la jambe portante. Des difficultés à produire ce freinage actif ont été observées chez les personnes âgées (Chong et al. 2009) et, à un degré plus important, chez les patients parkinsoniens (Chastan et al. 2009a) (Figure 13).

Avec l'avancée en âge, Chong et al. (2009) ont montré que le freinage actif du centre des masses est préservé pour une longueur de pas peu élevée ( $<0.4\text{m}$ ). Pour une longueur de pas plus importante ( $>0.4\text{m}$ ), ce mécanisme du contrôle postural est toujours observable mais est moins efficace. Ainsi, pour une même longueur de pas ( $>0.4\text{m}$ ), la vitesse verticale du centre des masses est plus élevée au moment du contact du pied oscillant sur le sol chez les personnes âgées par rapport à des sujets jeunes. Les réductions de longueur du pas et de vitesse de marche observées chez les personnes âgées lors de l'initiation de la marche (Muir et al. 2014) pourraient représenter des stratégies pour compenser ce déficit de contrôle du centre des masses (Chong et al. 2009).

Chastan et al. (2009a) ont montré que ce mécanisme du contrôle postural est aussi altéré chez des patients parkinsoniens, principalement à un âge avancé de la maladie et sans traitement dopaminergique. Ces auteurs ont aussi observé un lien entre l'altération du freinage vertical du centre des masses et la dégradation des capacités fonctionnelles des patients. Ce déficit de contrôle du centre des masses

pourrait être associé à une atrophie du mésencéphale (Chong et al. 2008 ; Demain et al. 2014). Il a aussi été proposé que les ganglions de la base, le cervelet et les afférences du système somatique sensoriel pourraient être impliqués dans la génération de ce mécanisme (Chastan et al. 2006, 2009a, b).



**Figure 13 : Exemples de tracés biomécaniques représentant la dynamique du centre des masses chez des sujets jeunes et âgés.** Tracés de gauche et du milieu (*Young*) : initiation de la marche chez un sujet jeune réalisant, respectivement, un pas court (0.33m) et long (0.63m). Tracés de droite (*Older*) : initiation de la marche chez une personne âgée avec un pas long (0.63m). TO (*toe off*) : décollement du pied oscillant, FC (*foot contact*) : contact du pied oscillant au sol. Vx : vitesse du centre des masses sur l'axe antéropostérieur, Vz : vitesse du centre des masses sur l'axe vertical, xP : déplacement du centre des pressions sur l'axe antéropostérieur. F (*forward*) : vers l'avant, B (*backward*) : vers l'arrière, + : vers le haut, - : vers le bas. V1 : vitesse maximale du centre des masses sur l'axe vertical lors de la phase d'exécution, V2 : vitesse du centre des masses sur l'axe vertical au moment du contact du pied oscillant sur le sol. Malgré une cinématique comparable, les sujets âgés ont une vitesse verticale du centre des masses plus élevée au moment du contact du pied oscillant sur le sol (V2) lorsque la longueur de pas et la vitesse de marche augmentent (tracés du milieu et de droite).

Leur capacité de freinage actif est plus faible que chez les jeunes. D'après Chong et al. (2009).

De façon intéressante, l'hypomobilité de la cheville induite par un port d'orthèse rigide peut aussi influencer la capacité de freinage vertical du centre des masses chez des jeunes adultes en bonne santé (Delafontaine et al. 2017). La modification des informations proprioceptives et l'incapacité à recruter correctement le triceps sural à cause de l'hypomobilité de la cheville ont probablement participé à la dégradation de ce freinage. Ces hypothèses pourraient s'appliquer aux personnes âgées ou aux patients parkinsoniens qui présentent des altérations des informations proprioceptives en distale (Gatev et al. 1999 ; Coquisart 2012) et une diminution des amplitudes articulaires de la cheville (Vandervoort et al. 1992 ; Halliday et al. 1998 ; Wu et al. 2020).

Enfin, il est à noter que le freinage actif du centre des masses lors de la phase d'exécution n'apparaît pas avant l'âge de 4 ans, ce qui suggère que ce mécanisme de contrôle de la force gravitationnelle nécessite un apprentissage par le système nerveux des contraintes posturales lors de la marche (Bril and Brenière 1992).

#### **1.4.5. Conclusion sur les effets de l'avancée en âge et de la maladie de Parkinson**

La prévalence des troubles de l'équilibre et de la marche augmente particulièrement avec l'avancée en âge (Pirker and Katzenschlager 2017). Ces problèmes accentuent le risque de chutes, qui ont parfois de graves conséquences pour la santé physique et mentale des personnes âgées. Les causes des troubles de la marche sont multiples. Elles peuvent être neurologiques (dysfonctionnements sensoriels ou moteurs, maladie de Parkinson etc.), orthopédiques (arthrose, arthrite, déformations osseuses etc.), médicales (insuffisances cardiaques, respiratoires, obésité etc.), psychologiques (peur de tomber, démence etc.) et enfin multifactorielles (surtout chez les personnes les plus âgées). Plus précisément, on retrouve chez la personne

vieillissante des altérations de la proprioception, une baisse de la vision, fréquemment de l'arthrose des hanches ou des genoux, une diminution des amplitudes articulaires et de la force musculaire, voire des lésions ou dysfonctionnements des structures cérébrales et spinales. Autrement dit, tous les systèmes qui interagissent pour le contrôle postural lors de l'initiation de la marche sont touchés par le vieillissement.

Chez les personnes âgées et les patients atteints de la maladie de Parkinson, les APA lors de l'initiation de la marche sont toujours présents. Toutefois, on observe souvent une baisse des déplacements habituels du centre des pressions, une augmentation de la durée de la phase posturale, voire des blocages chez les patients parkinsoniens. L'augmentation de la raideur chez ces populations pourrait aussi être responsable d'une dégradation du contrôle postural lors de la phase d'exécution. Cette instabilité dans le plan frontal lors de la phase unipodale, oblige les patients à effectuer un premier pas plus large, avec un moins bon contrôle de la pose du pied et des stratégies latérales de cheville. Enfin, il a été montré que les personnes âgées et les patients parkinsoniens pouvaient présenter un moins bon contrôle de la chute verticale du centre des masses lors de la phase d'exécution.

Dans la maladie de Parkinson, ces altérations peuvent en partie être contrecarrées par des traitements médicamenteux, notamment la Levodopa (Bryant et al. 2011). Néanmoins, ce traitement doit être utilisé à des doses minimales car il peut entraîner des complications à long terme, telles que des fluctuations de la réponse motrice et des dyskinésies, pouvant impacter la qualité de vie des patients (Simuni and Stern 1999 ; Tambasco et al. 2018). D'autres interventions non médicamenteuses peuvent avoir des actions préventives et thérapeutiques, comme les programmes d'activité physique par exemple (Ninot 2019). Certaines de ces interventions ont fait la preuve de leurs bénéfices sur la santé, la qualité de vie des patients, leur innocuité et viennent compléter les traitements conventionnels.

## **1.5. Effets de différents types d'exercices physiques sur la capacité à générer des APA**

De nombreux auteurs suggèrent que la perte de capacité à générer et à utiliser des APA constitue une cause majeure des troubles de l'équilibre (Mancini et al. 2009 ; Okada et al. 2011a, b ; Kanekar and Aruin 2014 ; Bonora et al. 2017). Il a été montré que les APA pouvaient être améliorés avec l'entraînement. Nous citerons ici différents exemples pour illustrer ce point.

Chez les patients parkinsoniens, un programme de renforcement des muscles des membres inférieurs et du tronc d'une durée de 10 semaines a permis d'améliorer le déplacement postérieur du centre des pressions lors des APA ainsi que la vitesse et la longueur de la foulée initiale (Hass et al. 2012). Ces résultats suggèrent qu'un programme adapté de musculation permet d'améliorer l'efficacité des APA et la performance motrice lors de l'initiation de la marche.

Les travaux de Marinho-Buzelli et al. (2017a, b) ont montré que lors de séances d'entraînement à la marche en balnéothérapie, la longueur et la vitesse de déplacement antéro-postérieur du centre des pressions lors des APA augmente par rapport à l'initiation de la marche sur le sol en surface. À l'inverse, la vitesse du centre des pressions est réduite lors de la phase d'exécution du premier pas. De plus, les forces de propulsion augmentent dans l'eau (afin de contrer les résistances à l'avancement) et l'amplitude active de la cheville augmente sans modification de la mobilité du genou (Marinho-Buzelli et al. 2017a, b). Ceci suggère que l'environnement aquatique pourrait faciliter l'entraînement à l'initiation de la marche pour les patients nécessitant une plus grande durée d'exécution (personnes âgées et patients parkinsoniens par exemple, Halliday et al. 1998) et ayant des difficultés à supporter ou transférer le poids du corps (après un AVC par exemple, Delafontaine et al. 2019b). Dans la même idée, l'ajout d'obstacles dans l'eau lors de rééducation en balnéothérapie chez des patients



parkinsoniens a permis d'améliorer encore plus significativement l'initiation de la marche, en diminuant notamment la fréquence des blocages (Zhu et al. 2018).

Vieira et al. (2015) ont montré que le chaussage et les propriétés physiques du sol affectent les déplacements du centre des pressions, notamment lors des APA sur l'axe médiolatéral. Sur une surface instable, le déplacement du centre des pressions sur l'axe médiolatéral (vers la jambe oscillante) est réduit, ce qui limite la propulsion du centre des masses du côté opposé. Autrement dit, il est plus difficile de transférer le poids du corps vers la future jambe portante. Ceci pourrait être lié au fait que les APA dépendent en grande partie de coordinations musculaires précises autour de l'articulation de la cheville (Crenna et al. 1991) et de la sensibilité plantaire (Lin et al. 2011). L'activité musculaire et les afférences sensorielles pourraient être affectées par la sensation d'instabilité sous les pieds (Vieira et al. 2015). Cet effet devrait être pris en compte et questionner l'utilisation des matériels instables en rééducation de la marche, notamment en début de progression.

Chez des personnes âgées, un entraînement régulier de 24 semaines comprenant des exercices d'initiation de la marche réalisés en double tâche (compter à rebours ou réciter l'alphabet) a permis d'améliorer le temps de réaction et le déplacement anticipateur vers l'arrière du centre des pressions lors de l'initiation de la marche en double tâche (Uemura et al. 2012b). Les difficultés à réaliser des activités en double tâche chez les personnes âgées et les patients parkinsoniens sont connues et certains auteurs suggèrent que les exercices en double tâche devrait être intégrés aux programmes de rééducation (Rochester et al. 2014).

Des entraînements prolongés en Tai Chi (48 semaines) chez des personnes âgées ont amélioré la génération des forces propulsives vers l'avant en augmentant le déplacement vers l'arrière du centre des pressions ainsi que la distance entre le centre des pressions et le centre des masses sur l'axe antéropostérieur lors des APA (Vallabhajosula et al. 2014). De plus, la coordination lors de l'initiation de la marche a aussi progressé après l'entraînement, ce qui suggère un meilleur contrôle postural (Hass et al. 2004).

Enfin, il a été montré qu'une session d'entraînement de lancers de *medecine ball* permet d'améliorer les APA lors de l'élévation bilatérale des membres supérieurs ou lors d'une perturbation externe prévisible chez des sujets jeunes (Kanekar and Aruin 2015), âgés (Aruin et al. 2015) ou atteints de sclérose en plaque (Aruin et al. 2017). Autrement dit, les muscles posturaux du tronc et des membres inférieurs étaient activés plus rapidement avant la perturbation, interne ou externe (Aruin et al. 2015). Ces résultats suggèrent que ce type d'entraînement qui se concentre sur les APA permet des adaptations transférables à d'autres activités et pourrait être efficace pour améliorer le contrôle postural, l'équilibre, la mobilité et la qualité de vie des sujets âgés. Néanmoins, les éventuels transferts des progrès sur le contrôle postural lors de la marche ou l'initiation de la marche n'ont pas encore été étudiés.

L'ensemble de ces travaux montre l'impact positif de programmes de rééducation sur la capacité des patients à générer des APA. Ces effets sont parfois méconnus des rééducateurs eux-mêmes. En effet, la plupart des programmes sont conçus de façon à améliorer les capacités de rattrapage des déséquilibres (ajustements posturaux correcteurs), alors que l'amélioration des capacités prédictives des patients peut s'avérer plus intéressante (Aruin et al. 2015 ; Aruin 2016)



# **Chapitre 2 : Problématique**

## 2. Problématique

Comme il a été souligné au cours de l'exposé du cadre théorique de cette thèse, la stabilité de l'équilibre est mise à l'épreuve lors de la marche et de l'initiation de la marche. Le système nerveux central est donc amené à développer un ensemble de mécanismes posturaux pour assurer une progression sécuritaire du corps dans la direction désirée et assurer la performance motrice (en terme de vitesse et/ou de longueur de pas). En accord avec le concept de capacité posturo-cinétique, chacun de ces mécanismes implique la mobilité de la chaîne posturale et en particulier celle de la cheville. Par conséquent, l'altération aigue ou chronique de cette mobilité, e.g. avec l'avancée en âge, aussi bien physiologique que pathologique (en particulier avec la maladie de Parkinson), peut potentiellement être source d'instabilité et/ou d'une diminution de la performance motrice. A l'inverse, une augmentation de cette mobilité, grâce à la pratique d'exercices physiques adaptés, pourrait permettre d'améliorer ces paramètres moteurs. C'est dans cette optique que l'effet sur la marche et l'initiation de la marche des VCE et d'étirements ciblés sur les muscles de la cheville ont été investigués dans différents types de population, saines ou pathologiques.

Les VCE sont de plus en plus répandues en rééducation auprès des personnes âgées avec des effets bénéfiques sur la force, l'endurance, la masse musculaire et l'équilibre (Orr 2015 ; Park et al. 2015 ; Rogan et al. 2017 ; Lam et al. 2018). Or, l'application des VCE est susceptible d'augmenter la raideur des membres inférieurs via un « réflexe tonique vibratoire » induit par une stimulation des fuseaux neuromusculaires (Hagbarth and Eklund 1966 ; Burke et al. 1976). Cette augmentation de la raideur, si elle était avérée dans notre étude, constituerait alors un facteur pouvant altérer la performance motrice et/ou la stabilité posturale, et ainsi exposer les sujets à un risque de chute accru (Yiou et al. 2016a). L'exposition aux VCE peut également entraîner des effets immédiats susceptibles d'altérer le développement des APA et le contrôle du positionnement du pied oscillant au sol. Ces mécanismes dépendent en

effet des informations sensorielles somatiques en provenance des membres inférieurs (Rugot et al. 2008, 2010 ; Saradjian et al. 2013 ; Roden-Reynolds et al. 2015). Or, il a été montré que celles-ci étaient altérées par l'exposition aux VCE (Pollock et al. 2010, 2011 ; Alfuth et al. 2012 ; Schlee et al. 2012). La dégradation de ces informations sensorielles, amorcée avec l'avancée en âge, pourrait donc être exacerbée par l'application des VCE, et ainsi avoir un effet négatif sur la stabilité et la performance motrice lors de la marche et l'initiation de la marche. Ainsi, la question de cette apparente contradiction entre ces effets potentiellement négatifs des VCE, et les effets positifs observés dans la littérature clinique reste posée.

A l'inverse des VCE, les étirements musculaires sont connus pour avoir des effets bénéfiques sur l'amplitude des mouvements articulaires, notamment autour de la cheville (Radford et al. 2006), et ce, après un programme de plusieurs semaines ou même une seule séance. En améliorant la mobilité de la chaîne posturale, les étirements musculaires sont donc susceptibles d'améliorer la capacité posturo-cinétique, et ainsi d'avoir un effet bénéfique sur la marche et l'initiation de la marche. Cet effet serait particulièrement profitable aux patients atteints de la maladie de Parkinson, chez qui la raideur des membres, associée à une hypertonie des muscles, constitue un symptôme très fréquent (Lee and Gilbert 2016 ; Tysnes and Storstein 2017 Raza et al. 2019 ; Wu et al. 2020).

**L'objectif général de cette thèse** est l'analyse systématique et expérimentale des effets d'une modification de la capacité posturo-cinétique induite par des VCE et des étirements musculaires sur l'organisation biomécanique de la marche et de l'initiation de la marche. Pour réaliser cet objectif, deux études expérimentales et deux revues systématiques de la littérature avec méta-analyses ont été réalisées.

Dans la première étude expérimentale, les effets immédiats des VCE sur l'organisation biomécanique de l'initiation de la marche ont été investigués chez des jeunes adultes et des personnes âgées saines. Dans la seconde étude expérimentale, les effets

immédiats d'une séance d'étirements ciblant spécifiquement le triceps sural sur l'organisation biomécanique de l'initiation de la marche ont été investigués chez des patients atteints de la maladie de Parkinson. Les hypothèses spécifiques à chacune de ces deux études sont détaillées dans les articles de la partie suivante (chapitre 3, présentation des travaux).

En parallèle de ces deux études expérimentales, deux revues systématiques de la littérature avec méta-analyses ont été réalisées : l'une portant sur les effets à long terme des VCE sur la marche et l'équilibre, et l'autre portant sur les effets à long terme des étirements sur la marche et l'équilibre, dans différents types de populations (saines et pathologiques).

Dans la suite du document, les effets des VCE et des étirements sont traités dans deux parties distinctes (« 3.1. Effets des vibrations corps entier sur la marche et l'initiation de la marche », et « 3.2. Effets des étirements musculaires sur la marche et l'initiation de la marche »). Les résultats principaux sont synthétisés dans le dernier chapitre du document, où nous nous proposerons des applications pratiques et des perspectives de recherche. Enfin, nous présentons en annexe un article (revue systématique) qui ouvre les perspectives de nos travaux vers le domaine des patients hémiplésiques.

# **Chapitre 3 : Présentation des travaux**



### **3. Présentation des travaux**

#### **3.1. Effets des vibrations corps entier sur la marche et l'initiation de la marche**

##### **3.1.1. Effets à long terme des vibrations corps entier sur la marche : revue systématique de la littérature et méta-analyse**

Cette étude a donné lieu à la publication suivante :

Fischer M, **Vialleron T**, Laffaye G, Fourcade P, Hussein T, Chèze L, Deleu PA, Honeine JL, Yiou E, Delafontaine A (2019). Long-Term Effects of Whole-Body Vibration on Human Gait: A Systematic Review and Meta-Analysis. *Front Neurol* 10: 627. Doi: 10.3389/fneur.2019.00627.



# Long-Term Effects of Whole-Body Vibration on Human Gait: A Systematic Review and Meta-Analysis

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**Background:** Whole-body vibration is commonly used in physical medicine and neuro-rehabilitation as a clinical prevention and rehabilitation tool. The goal of this systematic review is to assess the long-term effects of whole-body vibration training on gait in different populations of patients.

**Methods:** We conducted a literature search in PubMed, Science Direct, Springer, Sage and in study references for articles published prior to 7 December 2018. We used the keywords “vibration,” “gait” and “walk” in combination with their Medical Subject Headings (MeSH) terms. The Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) methodology was used. Only randomized controlled trials (RCT) published in English peer-reviewed journals were included. All patient categories were selected. The duration of Whole-Body Vibration (WBV) training had to be at least 4 weeks. The outcomes accepted could be clinical or biomechanical analysis. The selection procedure was conducted by two rehabilitation experts and disagreements were resolved by a third expert. Descriptive data regarding subjects, interventions, types of vibration, training parameters and main results on gait variables were collected and summarized in a descriptive table. The quality of selected studies was assessed using the PEDro scale. Statistical analysis was conducted to evaluate intergroup differences and changes after the WBV intervention compared to the pre-intervention status. The level of evidence was determined based on the results of meta-analysis (effect size), statistical heterogeneity ( $I^2$ ) and methodological quality (PEDro scale).

**Results:** A total of 859 studies were initially identified through databases with 46 articles meeting all of the inclusion criteria and thus selected for qualitative assessment. Twenty-five studies were included in meta-analysis for quantitative synthesis. In elderly subjects, small but significant improvements in the TUG test (SMD =  $-0.18$ ; 95% CI:  $-0.32$ ,  $-0.04$ ) and the 10MWT (SMD =  $-0.28$ ; 95% CI:  $-0.56$ ,  $-0.01$ ) were found in the WBV groups with a strong level of evidence ( $I^2 = 7\%$ ,  $p = 0.38$  and  $I^2 = 22\%$ ,  $p = 0.28$ , respectively; PEDro scores  $\geq 5/10$ ). However, WBV failed to improve the 6MWT (SMD =  $0.37$ ; 95% CI:  $-0.03$ ,  $0.78$ ) and the Tinetti gait scores (SMD =  $0.04$ ; 95% CI:  $-0.23$ ,  $0.31$ )

in older adults. In stroke patients, significant improvement in the 6MWT (SMD = 0.33; 95% CI: 0.06, 0.59) was found after WBV interventions, with a strong level of evidence ( $I^2 = 0\%$ ,  $p = 0.58$ ; PEDro score  $\geq 5/10$ ). On the other hand, there was no significant change in the TUG test despite a tendency toward improvement (SMD =  $-0.29$ ; 95% CI:  $-0.60, 0.01$ ). Results were inconsistent in COPD patients ( $I^2 = 66\%$ ,  $p = 0.03$ ), leading to a conflicting level of evidence despite a significant improvement with a large effect size (SMD = 0.92; 95% CI: 0.32, 1.51) after WBV treatment. Similarly, the heterogeneous results in the TUG test ( $I^2 = 97\%$ ,  $p < 0.00001$ ) in patients with knee osteoarthritis make it impossible to draw a conclusion. Still, adding WBV treatment was effective in significantly improving the 6 MWT (SMD = 1.28; 95% CI: 0.57, 1.99), with a strong level of evidence ( $I^2 = 64\%$ ,  $p = 0.06$ ; PEDro score  $\geq 5/10$ ). As in stroke, WBV failed to improve the results of the TUG test in multiple sclerosis patients (SMD =  $-0.11$ ; 95% CI:  $-0.64, 0.43$ ). Other outcomes presented moderate or even limited levels of evidence due to the lack of data in some studies or because only one RCT was identified in the review.

**Conclusions:** WBV training can be effective for improving balance and gait speed in the elderly. The intervention is also effective in improving walking performance following stroke and in patients with knee osteoarthritis. However, no effect was found on gait quality in the elderly or on balance in stroke and multiple sclerosis patients. The results are too heterogeneous in COPD to conclude on the effect of the treatment. The results must be taken with caution due to the lack of data in some studies and the methodological heterogeneity in the interventions. Further research is needed to explore the possibility of establishing a standardized protocol targeting gait ability in a wide range of populations.

**Keywords:** whole-body vibration, long-term effects, gait, biomechanics, randomized controlled trials, meta-analysis

## HIGHLIGHTS

- WBV is currently used in locomotor rehabilitation.
- WBV presents strong evidence for improving performance in the timed-up-and-go test in the elderly, but not in stroke or multiple sclerosis patients.
- WBV presents strong evidence for improving performance in the 10-meter walk test for elderly, in the 6-min walk test for stroke and knee OA patients but results are conflicting in COPD patients.
- Other outcomes present moderate or limited levels of evidence due to the lack of data or because only one RCT was identified in other pathologies.

## INTRODUCTION

Whole-body vibration (WBV) is a therapeutic method that exposes the entire body to mechanical oscillations while the patient stands or sits on a vibrating platform. This method was first used in the late nineteenth century by Charcot to treat gait disorders in neurological patients, especially in patients with Parkinson's disease (1). It is now commonly used in the physical medicine/neuro-rehabilitation fields as a prevention and rehabilitation tool for sarcopenia (2), osteoporosis (3), chronic low back pain (4), and fibromyalgia (5), among other conditions.

WBV is also used in rehabilitation to improve muscle function (strength, power, and endurance) (6), muscle soreness (7), joint stability (8) and to reduce the risk of falling (9).

Several spinal and supraspinal mechanisms have been proposed to explain increased muscle activity during exposure to WBV. While there is currently no consensus, the most frequently cited mechanism is a reflex muscular contraction called tonic vibration reflex (TVR). This phenomenon has been shown to occur during direct and indirect vibratory musculo-tendinous stimulations that excite muscle spindles and enhance activation of Ia afferents, resulting in a higher recruitment of motor units and gradual development of muscle activity (10). In addition to these spinal reflexes, neuromuscular changes (11, 12), increased intramuscular temperature (10) and peripheral blood flow (13) may contribute at different levels to the increased muscular performance observed after WBV.

A recent review (14) reported a beneficial effect of long-term WBV training on balance control under static postural conditions. Since the literature appears to suggest a neuroanatomic (15) and a biomechanical continuum between standing posture and gait (16–18), Rogan et al. suggested that this beneficial effect could be extended to dynamic motor tasks such as gait (14). Such a continuum has been analyzed in stroke patients (19), for example. The most recent literature review focusing on the effect of WBV on gait, however, provided

only mitigated support for this assumption (20). Based on the screening of 10 randomized controlled trials (RCT), Lindberg and Carlsson concluded there was low-quality evidence for the beneficial use of long-term WBV on gait, and acknowledged there were major limitations (20), the most important being that only one of the authors reviewed the literature. Thus, no group discussions were conducted with experts to resolve possible disagreements and reach a mutual consensus. In addition, the low number of RCT included ( $n = 10$ ) and the absence of meta-analysis may have limited the relevance of Lindberg and Carlsson's review. Since that review was published, WBV training has been used increasingly in physiotherapy to prevent and/or treat gait disorders. Consequently, more and more experimental studies have been conducted in this area with both healthy and pathological participants.

Hence, the aim of this article is to provide an up-to-date literature review of RCT studies on the effects of long-term WBV training on gait in both healthy subjects and pathological patients. It will contribute to provide evidence-based practice for a promising non-pharmacological rehabilitative method that is both safe and cheap, and that can be used by patients at home as part of an auto-rehabilitation program.

## MATERIALS AND METHODS

### Design and Literature Screening

The Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) methodology was employed in this systematic review (Figure 1).

The PubMed, Science Direct, Springer and Sage databases were used for a comprehensive systematic literature search for articles published prior to 7 December 2018 with no time limit. The keywords used were: "vibration" AND (gait OR walk). More specifically, the search details specified in PubMed were: ("vibration"[MeSH Terms] OR "vibration"[All Fields]) AND (("gait"[MeSH Terms] OR "gait"[All Fields]) OR ("walking"[MeSH Terms] OR "walking"[All Fields] OR "walk"[All Fields])).

The selection procedure was conducted by two experts in rehabilitation. Disagreements were discussed with a third expert in a group until a mutual consensus was reached. First, a review was performed on all available titles obtained from the literature search with the selected keywords. All relevant or potentially relevant titles were included in the subsequent phase. Then, the abstracts were reviewed with all relevant or potential articles included in the following phase. Finally, full-text articles were reviewed to ensure that only relevant studies were included. In the same way, reference lists of all included articles were reviewed to possibly include articles through cross-referencing.

### Inclusion and Exclusion Criteria

To be included, the studies had to meet all of the following inclusion criteria: all patient categories were selected if: gait ability was measured before and after at least 4 weeks of WBV training performed on a vibration platform; the results were based on biomechanical analyses or were clinically relevant;

the control group had no intervention or performed the same physical rehabilitation, resistance, balance or endurance training as the intervention group. In addition, only RCT, articles in English, and articles published in peer-reviewed journals were included. Studies were excluded if they measured only short-term effects ( $< 4$  weeks) and if WBV was combined with non-physical training or with any intervention not provided to the control group (i.e., not only WBV effects are measured).

### Data Extraction and Main Measurements Examined

Data were extracted from the selected articles by one of the authors. The extracted data were checked by another author and disagreements were resolved with a third.

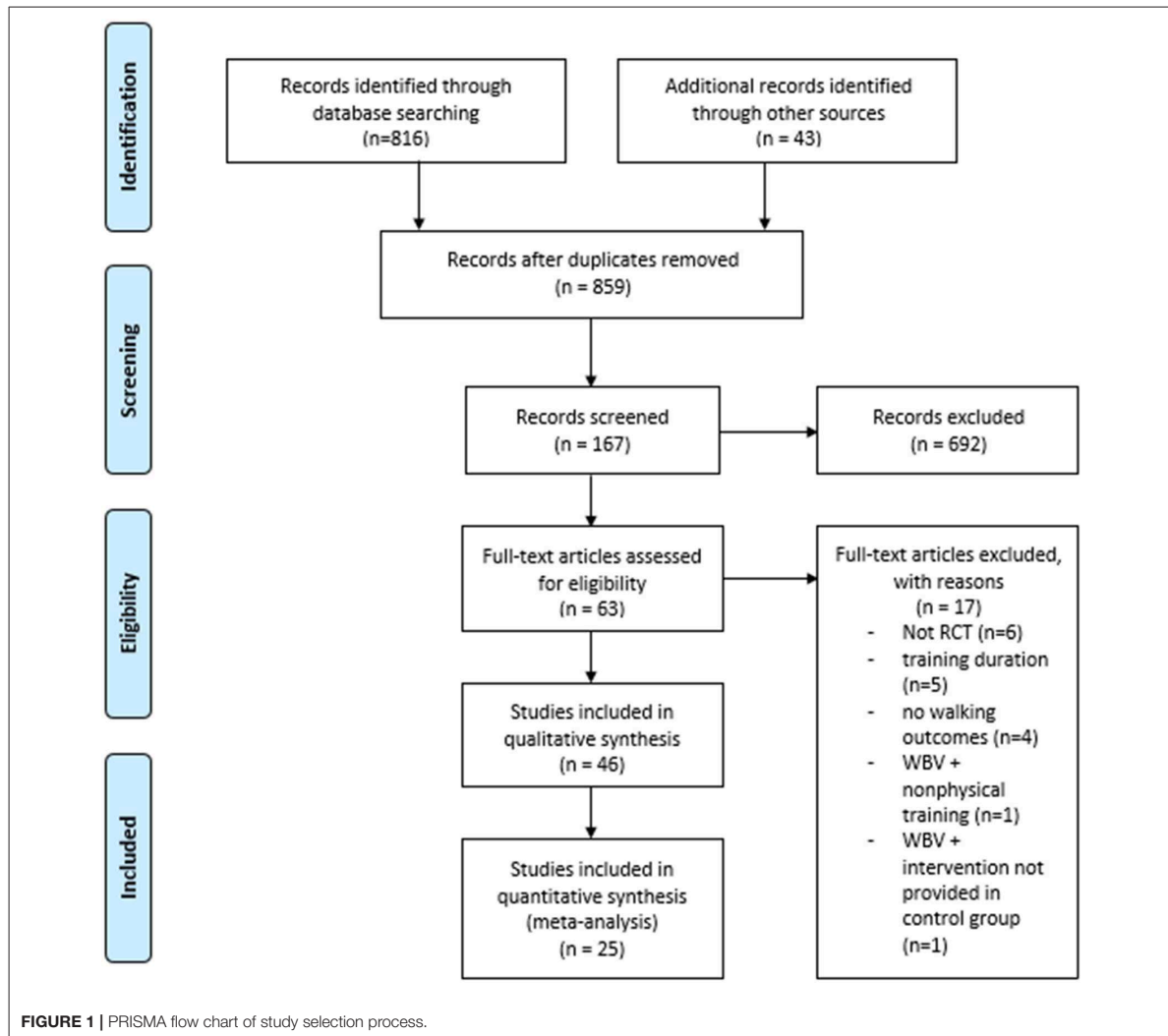
The following data were extracted for each selected article: (1) the names of the authors and the date of publication; (2) the number of subjects involved in the experiment with their characteristics and breakdown in each group; (3) WBV training details (in the following order: name of the WBV device, duration of the intervention, number of sessions, types of exercises, number of vibration sets, exposure duration per set, rest period between sets, frequency, amplitude and type of vibration) and control group details; and (4) the main outcomes related to gait with the main results (e.g., timed up-and-go test, 6-min walk test, walking speed, etc). When information could not be provided, it was indicated by a "?".

### Quality and Risk of Bias Assessment

The PEDro scale was used to assess the risk of bias, and thus the methodological quality of the selected studies (21). The scale was chosen for its ability to provide an overview of the external (criterion 1), internal (criteria 2–9) and statistical (criteria 9 and 10) validity of RCT. The scale is divided in 11 criteria, but the first criterion is not calculated in the total score. The output of each criterion could be either "yes" (y), "no" (n) or "do not know" (?). A "y" was given a score of one point, while a "n" or "?" was assigned zero points. Studies with a total score of 5–10/10 ( $\geq 50\%$ ) were considered to be of high quality, and scores of 0–4/10 ( $<50\%$ ) as low quality (20). Two evaluators assessed the quality of the included studies independently. In the event of disagreements, a group discussion was held with a third expert to reach a mutual consensus.

### Statistical Analysis

To estimate the effect of WBV training on human gait, a meta-analysis compared the intervention groups with the control groups. Within group comparisons were added (i.e., pre vs. post intervention) when the groups were not comparable (e.g., statistical difference in outcomes at baseline or additional training in control group not provided in the intervention group). Estimations were calculated using the methodology described by Wan et al. (22) when mean and standard deviations were not reported by the authors and medians and interquartile ranges were used. The authors were contacted to request additional data when an estimation was not possible.



If no response was received, the variables were excluded from meta-analysis.

Statistical analysis and figures (i.e., forest plot to facilitate the visualization of values) were produced using a random-effect model in Review Manager software (RevMan, v 5.3, Cochrane Collaboration, Oxford UK) (23). A random-effect model was used to take into account the heterogeneity between the study effects. The effect size of the interventions was reported by standard mean difference (SMD) and their respective 95% Confidence Interval (CI). In this way, the magnitude of the overall effect can be quantified as very small ( $<0.2$ ) small (0.2–0.49), moderate (0.5–0.79) or large ( $\geq 0.8$ ) (24, 25). Statistical heterogeneity was calculated using the  $I^2$  and Cochrane Q statistic tests (25). Statistical significance was set at  $p < 0.05$ .

### Level of Evidence

The strength of evidence of primary outcomes was established as described by Van Tulder et al. (26) based on the results of meta-analysis (effect size), statistical heterogeneity ( $I^2$ ) and risk of bias (PEDro scale). The level of evidence was considered strong with multiple high-quality RCT (at least two studies with a PEDro score  $\geq 5/10$ ) that were statistically homogenous ( $I^2 p \geq 0.05$ ). The level of evidence was considered moderate with multiple low-quality studies (two studies with a PEDro score  $< 5/10$ ) that were statistically homogenous and/or one high quality RCT. The level of evidence was considered limited when only one low quality RCT was identified. The level of evidence was conflicting when there were multiple statistically heterogenous studies ( $I^2 p < 0.05$ ).

## RESULTS

### Included Studies

A total of 816 titles were screened in the first search stage, 43 more were included through cross-referencing, and 692 were excluded because they did not concern our research question. The main reasons for exclusion were: absence of WBV treatment (e.g., studies using local vibrations were excluded), measurement of acute effects, no value for dynamic balance, case studies and reviews. Following exclusion, 167 studies were considered for an abstract review. A further 104 were excluded in this second stage because they did not meet the inclusion criteria. Finally, 63 full-text articles were assessed for eligibility with 17 not accepted: five because training lasted < 4 weeks, six because they were not RCT, four because there were no walking outcomes, one because it combined WBV training with non-physical therapy and one for comparing WBV training combined with another intervention not provided in the control group (meaning that not only WBV effects were measured). Thus, 46 articles were ultimately included in this systematic review (9, 27–69, 71, 72). A summary of the study selection is provided in **Table 1**.

### Characteristics of the Populations

A total of 2 029 patients took part in the 46 studies selected in this review (see **Table 1**). The sample size ranged from 14 to 159 participants, with a mean age of  $60.9 \pm 20.0$  years, varying from 7.9 years to 83.2 years. With regard to the adult population, 16 studies evaluated the effects of WBV in the elderly ( $n = 59.8 \pm 35.4$  subjects) (9, 30–32, 36, 40, 44, 45, 50, 53, 55, 56, 59, 62, 64, 69), four in patients with Chronic Obstructive Pulmonary Disease (COPD) ( $n = 42.5 \pm 16.7$  subjects) (57, 61, 65, 71), seven in patients with stroke ( $n = 46.1 \pm 27.2$  subjects) (28, 35, 38, 51, 54, 67, 72), four in patients with osteoarthritis (OA) ( $n = 32.2 \pm 11.9$  subjects) (29, 33, 63, 68), three in postmenopausal women ( $n = 40.3 \pm 12.5$  subjects) (48, 58, 66), two in patients with multiple sclerosis ( $n = 29.5 \pm 6.3$  subjects) (34, 39) and one in patients with the following pathologies: incomplete cervical spinal injury (47), pulmonary arterial hypertension (42), lung transplantation (43), idiopathic Parkinson's disease (41), total knee arthroplasty (49) and cerebral palsy (27) ( $n = 30.0 \pm 26.4$  subjects). With regard to the child population, two studies evaluated the effects of WBV in cerebral palsy (37, 60), one in patients with osteogenesis imperfecta (46) and one in patients with spastic diplegia or quadriplegia forms of cerebral palsy (52) ( $n = 22.5 \pm 5.9$  subjects). Most of the studies included both males and females, except for nine studies that either did not mention the participants' gender or selected only males or females (including the three studies on post-menopausal women). Most of the studies clearly explained their eligibility criteria and had similar baselines (no significant differences between groups in any outcomes before the intervention) in their groups, except in 10 articles.

### Training Protocols

The duration of the WBV training interventions ranged from four to 32 weeks, with between two and five sessions per week, with a mean of  $3.1 \pm 0.8$  (three sessions per week in 31 of the 46 selected articles). The frequency and amplitude

used in the training sessions ranged from 2 to 45 Hz and from 0.44 to 20 mm, respectively. The intensity of the training sessions, by frequency and/or amplitude, was progressively increased in 30 studies, and remained unchanged in the other selected studies. Some WBV platforms delivered the vibrations alternating between the right and the left foot, while the right and left foot moved up and down at the same time in other vibration plates (70). Synchronous vibrations were delivered in 20 studies, side-alternating vibrations were used in 11 studies, while 15 studies did not mention the type of vibration in their intervention method.

For the groups that were exposed to WBV training (interventions groups), vibrations were delivered while participants stood in static positions (e.g., squat or lunge positions) in 27 studies and dynamic exercises were provided in 11 studies. In the remaining eight studies, both static and dynamic exercises were combined during the WBV training sessions. The number of WBV sets per training session ranged between 1 and 135. The duration of the vibration sets ranged from 10 s to 3 min, with a between-sets resting time ranging between 3 s and 5 min. For the groups not exposed to WBV training interventions (control groups), participants performed strengthening and balance exercises without WBV in fourteen studies, had no intervention and were asked to maintain their habitual lifestyle in sixteen studies, were exposed to a sham intervention in six studies, continued to follow their conventional physiotherapy in four studies, received relaxation exercises in four studies and performed walking training sessions in two studies.

### Gait Motor Outcomes

The “Timed Up-and-Go” (TUG) test and the “six-minute walking test” (6MWT) were the clinical outcomes most frequently used to assess gait (in 29 and 18 studies, respectively). The “ten-meter walking test” (10MWT) was used in 10 studies to assess gait velocity. Walking speed was also evaluated using biomechanical and kinematic assessments (e.g., walking on a platform or camera motion analysis) in six studies. Other temporal and spatial parameters such as time of swing phase and stance phase, stride length and step length were presented in only two studies. Gait quality was assessed using the gait score of the Tinetti test in five studies. Finally, other outcomes were used once in all 46 studies: the “functional ambulation categories test” with stroke patients, the “50-foot walking test” with knee OA patients, the “25-foot walking test” with multiple sclerosis patients, the “two-minute walking test” with knee OA patients, and the time to walk four meters in postmenopausal women. A summary of the primary outcomes related to gait is provided in **Table 1**.

### Quality Assessment

The results from the quality assessments for each of the studies for respective quality indexes are provided in **Table 2**. According to the PEDro Scale, 40 studies obtained a high-quality methodology score while six studies were rated as low quality.

The mean score was  $5.8 \pm 1.4$  with a median of 5.5 and a range of scores from 3 to 9. The highest-quality methodology scores were found in the articles concerning stroke patients, with a mean

**TABLE 1 |** Descriptive checklist of the included studies.

Article	Subjects	Interventions	Outcomes (only intergroup differences are presented)
<b>OLDER ADULTS</b>			
Lam et al. (50)	73 older adults, 40 women, mean age: 82.3 ± 7.3 years. WBV + exercise (WBV+E): <i>n</i> = 25, 12 men, 13 women, mean age 84 years Exercise: <i>n</i> = 24, 10 men, 14 women, mean age 82.4 years. Control: <i>n</i> = 24, 11 men, 13 women, mean age 80.3 years.	8 weeks, 3 times per week. WBV + E: Fitvibe medical WBV system (GymnaUniphy NV, Bilzen, Belgium), dynamic exercises, 4 × 1 min/1–2 min, 30 Hz (weeks 1–4) and 40 Hz (weeks 5–8), 0.9 mm, vertical displacements. Exercise: identical exercise program without WBV. Control: social and recreational activities that only involved the upper limbs.	WBV + E vs. Exercise: No significant effect for TUG (SMD = −0.18, 95% CI: −0.74, 0.38) and the 6MWT (SMD = 0.21, 95% CI: −0.35, 0.78).
Wei et al. (69)	80 community dwelling seniors with sarcopenia. Low frequency group: <i>n</i> = 20, 7 men, 13 women, mean age 78 years Medium frequency group: <i>n</i> = 20, 7 men, 13 women, mean age 75 years High frequency group: <i>n</i> = 20, 5 men, 15 women, mean age 74 years Control group: <i>n</i> = 20, 5 men, 15 women, mean age 76 years	3 days/week over a 12-week period, WBV: 4 mm for all training groups, knee joint flexed at 60°, Fitvibe excel, GymnaUniphy NV, Bilzen, Belgium, vertical vibrations. Low frequency group: 20 Hz × 720 s Medium frequency group: 40 Hz × 360 s High frequency group: 60 Hz × 240 s Control group: no extra training	Low frequency group vs. control: no significant difference for the TUG test (SMD = −0.22, 95% CI: −0.84, 0.41) Medium frequency group vs. control: no significant difference for the TUG test (SMD = −0.40, 95% CI: −1.03, 0.22) High frequency group vs. control: no significant difference for the TUG test (SMD = −0.30, 95% CI: −0.92, 0.33)
Goudarzian et al. (45)	42 healthy old men. WBV: <i>n</i> = 11, mean age 66, 58 years. MT: <i>n</i> = 12, mean age 69, 20 years. WBV+MT: <i>n</i> = 10, mean age 67, 80 years. Control: <i>n</i> = 9, mean age 68, 90 years.	3 times a week, 8 weeks WBV: Novotec, Pforzheim, Germany, static and dynamic exercises, 6 × 45–85s/45–85s, 30–35 Hz, 5–8 mm, <i>n</i> MT: relaxation techniques. WBV+MT: combination of vibration and MT that was the half-time of each protocol. Control: Daily routine.	WBV vs. Control: no difference between group for the TUG test (SMD = −0.60, 95% CI: −1.50, 0.31). Significant improvement of the 10MWT in favor of the WBV (SMD = −1.32, 95% CI: −2.32, −0.33).
Sitjā-Rabert et al. (64)	159 older people, 107 women, 52 men, with a mean age of 82 years. WBV + exercise group: <i>n</i> = 81, <i>n</i> Exercise group: <i>n</i> = 78, <i>n</i>	6 weeks, 3 sessions per week. WBV + exercise group: Pro5 Airdaptive Model; PowerPlate, Amsterdam, The Netherlands, static/dynamic exercises, <i>n</i> , 30–35 Hz, 2–4 mm, <i>n</i> Exercise group: same static/dynamic exercises without vibration platform.	WBV + exercise group vs. exercise group: No difference between group for the TUG test (SMD = −0.02, 95% CI: −0.39, 0.34) No difference between group for the Tinetti gait score (SMD = −0.08, 95% CI: −0.44, 0.27)
Santin-Medeiros et al. (62)	37 elderly women, mean age 82.4 years. WBV group: <i>n</i> = 25. Control group: <i>n</i> = 18.	8-month, 2 sessions per week WBV: Fitvibe Excel Pro; GymnaUniphy NV Bilzen, Belgium, 18 exercises, 6/session, 1–2 sets/exercise, 30–35/exercise, 6 min–6 min50 s/ session, 20 Hz, 2 mm. Control: maintain their habitual lifestyle	WBV vs. control: Groups were statistically different at baseline for the TUG test. WBV: No significant improvement of the TUG test post WBV (SMD = 0.15, 95% CI: −0.49, 0.78).
Buckinx et al. (36)	62 nursing home residents. WBV group: <i>n</i> = 31, 11 men, 20 women, mean age 82.2 years. Control group: <i>n</i> = 31, 3 men, 27 women, mean age 84.2 years.	6 months, 3 training sessions every week. WBV group: Vibrosphère, knees flexed, 5 × 15 s/30 s, 30 Hz, 2 mm, vertical vibrations. Control group: normal daily life.	Lack of data post WBV. WBV vs. control: authors reported no significant inter group difference for the TUG test, Tinetti gait score and for the parameters recorded by the Locometrix ( <i>p</i> > 0.05).
Lee et al. (53)	55 Elderly Patients with Diabetic Neuropathy WBV + BE group: <i>n</i> = 19, 9 men, 10 women, mean age 76.31 years. Balance exercise group (BE): <i>n</i> = 18, men:7, women:11, mean age: 74.05 years. Control group: <i>n</i> = 18, 8 men, 10 women, mean age 75.77 years.	6 weeks, twice per week, same physical therapy. WBV + BE group: Galileo 2000, Novotec Medical GmbH, Germany, 3/week, squatting position, 3 × 3 min/1-min, 15–30 Hz, 1–3 mm, <i>n</i> BE group: strength, balance, and functional mobility training. Control group: <i>n</i>	WBV + BE group vs. BE group: Significant improvement of the TUG test in favor of the WBV group (SMD = −0.72, 95% CI: −1.39, −0.06).

(Continued)

TABLE 1 | Continued

Article	Subjects	Interventions	Outcomes (only intergroup differences are presented)
Beaudart et al. (31)	62 nursing home residents. WBV group: $n = 31$ , 11 men, 20 women, mean age 82.2 years. Control group: $n = 31$ , 4 men, 27 women, mean age 84.2 years.	3 months, 3 training sessions every week. WBV: Vibrosphere, static position with a knee flexion, $5 \times 15/30$ s, 30 Hz, 2 mm, vertical vibrations. Control group: requested neither to change their lifestyle during the study nor to get involved in any new type of physical activity.	WBV vs. control: No significant difference between groups for the TUG test (SMD = $-0.10$ , 94% CI: $-0.59$ , 0.40) and the Tinetti test (SMD = $0.30$ , 95% CI: $-0.20$ , 0.80). Lack of data post WBV for the Locomotrix system. The authors reported no significant intergroup difference for the parameters recorded by the Locomotrix ( $p > 0.05$ ).
Gómez-Cabello et al. (44)	49 non-institutionalized elderly (20 men and 29 women; aged $75.0 \pm 4.7$ years). WBV: $n = 24$ , $n$ Control: $n = 25$ , $n$	11 weeks, 3 times per week. WBV: Pro5 Power plate, London, UK, squat position, $10 \times 45/60$ s, 40 Hz, 2 mm. Control: not participate in any training. Asked not to change their lifestyle.	WBV vs. control: No difference between group for the 6MWT (SMD = $0.54$ , 95% CI: $-0.03$ , 1.11).
Bogaerts et al. (32)	111 elderly women over 70 years of age WBV group: $n = 54$ Control group: $n = 57$	6 months, 3 sessions per week; WBV group: Powerplate, 2–5 dynamic exercises. $4 \times 15$ s/60s (start of the study), $12 \times 60$ s/5s (6months). 30-40HZ, 1,6–2,2g, $n$ Control group: no training program.	WBV vs. control group: No significant difference for the TUG test (SMD = $-0.18$ , 95% CI: $-0.55$ , 0.20) and the 10MWT (SMD = $-0.26$ , 95% CI: $-0.63$ , 0.12) at preferred speed. No significant difference for the TUG test (SMD = $-0.31$ , 95% CI: $-0.68$ , 0.07) and the 10MWT (SMD = $-0.10$ , 95% CI: $-0.47$ , 0.27) at maximum speed.
Mikhael et al. (56)	19 older adults mean age 64, 4 years (range 50–80). WBV with flexed knees (FK): $n = 6$ , 4 men, 2 women, mean age 63.3 years WBV with locked knees (LK): $n = 5$ , 3 men, 2 women, mean age 69 years Sham: $n = 8$ , 4 men, 4 women, mean age 62.3 years	20 min, 3 days per week, 3 months, static exercises, $39 \times 1$ min/1min, 12 Hz, 1 mm WBV with FK: vibration platform engineered by Australian Catholic University (2004), knee angle at 20 WBV with LK: lock knees Sham: The amplitude was set to 0 mm, giving 0g magnitude.	Lack of data post WBV. The authors reported no between groups difference after WBV for the 6MWT ( $p = 0.61$ ), habitual and maximal gait velocities ( $p = 0.80$ and $p = 0.58$ , respectively).
Machado et al. (55)	26 community-dwelling elderly women WBV: $n = 13$ , mean age 79.3 years Control: $n = 13$ , mean age 76.2 years	3–5 times a week, 10 weeks WBV: Fitvibe, GymnaUniphy NV, Bilzen, Belgium, half squat, deep squat, wide stance squat, calves, 1–2 sets/exercise, 30–60 s/120–180 s, 2–4 mm, 20–40 Hz, increased progressively, $n$ Control: requested to do not change their lifestyle during the study	Lack of data post WBV. The authors reported a significant improvement of the TUG test post WBV ( $p < 0.05$ ) but no significant difference between groups ( $p > 0.05$ ).
Furness and Maschette. (40)	73 older adults, 38 females and 35 males, mean age $72 \pm 8$ years 1 WBV session per week: $n$ 2 WBV sessions per week: $n$ 3 WBV sessions per week: $n$ Control group: $n$	0, 1, 2, or 3 times a week, 6 weeks. WBV interventions: $n$ , static, knees flexes at $110^\circ$ , $5 \times 1$ min/1min, 15–25 Hz, 0,5mm, vertical vibrations. Control group: The zero group did not participate in any WBV sessions.	1 WBV vs. control: No significant between group difference for the TUG test (SMD = $0.57$ , 95% CI: $-0.10$ , 1.24) 2 WBV vs. control: No significant between group difference for the TUG test (SMD = $0.57$ , 95% CI: $-0.10$ , 1.24) 3 WBV vs. control: No significant between group difference for the TUG test (SMD = $-0.45$ , 95% CI: $-1.11$ , 0.20)
Rees et al. (59)	43 older adults, untrained, 23 men and 20 women WBV group: $n = 15$ , mean age 74.5 years Exercise group: $n = 13$ , mean age 73.1 years Control group: $n = 15$ , mean age 73.1 year	3 sessions a week, 8weeks, low-intensity walking at least 3 times a week WBV group: Novotec, Pforzheim, Germany, static and dynamic exercises, $6 \times 45$ –80s/45–80s, 26 Hz, 5–8 mm, increased progressively, vertical displacements. Exercise group: same exercises without WBV. Control group: low intensity walking	WBV vs. exercise group: No significant difference between groups for the TUG test (SMD = $-0.35$ , 95% CI: $-1.10$ , 0.40) and the 10MWT (SMD = $-0.25$ , 95% CI: $-0.99$ , 0.50). WBV vs. control: no significant difference between group for the TUG test (SMD = $-0.22$ , 95% CI: $-0.75$ , 0.31) and the 10MWT test (SMD = $-0.29$ , 95% CI: $-1.01$ , 0.43).

(Continued)



TABLE 1 | Continued

Article	Subjects	Interventions	Outcomes (only intergroup differences are presented)
Bautmans et al. (30)	24 older adults, nursing home residents. WBV: $n = 13$ , 5 men, 8 women, mean age 76.6 years. Sham: $n = 11$ , 4 men, 7 women, mean age 78.6 years.	3 times weekly, 6 weeks WBV: Power- Plate, Badhoevedorp, The Netherlands, 2–4 static lower limb exercises/sessions, 1–3 × 30–60s/30–60s, 35–40 Hz, 2–5 mm, increased progressively, vertical vibrations. Sham: same exercise program on the vibration plate, but without vertical vibrations.	WBV vs. sham: No significant difference between groups for the TUG test (SMD = $-0.38$ , 95% CI: $-1.25$ , $0.48$ ) and the Tinetti test (SMD = $0.00$ , 95% CI: $-0.86$ , $0.86$ ).
Bruyere et al. (9)	42 older adults, nursing home residents WBV + Physical therapy: $n = 22$ , 4 men, 18 women, mean age 84.5 years Physical therapy: $n = 20$ , 7 men, 13 women, mean age 78.9 years	3 times a week, 6 weeks, same PT 10 min WBV + Physical therapy: $n$ static exercise, 4 × 60 s/90 s, 10–26 Hz, 3–7 mm, vertical vibrations. Physical therapy: PT only.	Lack of data post WBV. Groups were statistically different at baseline for the TUG test ( $p = 0.04$ ). The authors reported a significant decrease of $11.0 \pm 8.6$ s post WBV for the TUG test and an increase of $3.5 \pm 2.1$ points post WBV for the Tinetti gait score.
<b>PATIENTS WITH CHRONIC OBSTRUCTIVE PULMONARY DISEASE (COPD)</b>			
Spielmanns et al. (71)	28 subjects with COPD stage II-IV. WBVT group: $n = 12$ , 8 men, 4 women, mean age 62.4 years. Conventional training group (CTG): $n = 16$ , 9 men, 7 women, mean age 68 years.	3 months, 2 sessions/week, same resistance, and endurance training. WBVT group: Galileo vibration platform (No- votec Medical, Pforzheim, Germany), 3 × 20 squat repetitions, 24–26 Hz, 3 mm, side-alternating vibration. CTG: same of squat exercises but without WBVT.	WBV vs. control: no significant difference between groups for the 6MWT (SMD = $0.72$ , 95% CI: $-0.05$ , $1.50$ ).
Spielmanns et al. (65)	29 subjects with stable COPD in stage I to III WB group: $n = 14$ , 7 men, 7 women, mean age 69 years. Calisthenics group: $n = 14$ , 7 men, 7 women mean age 70 years.	3 months, twice per week. WBV group: Galileo, Novotec Medical, Pforzheim, Germany, isometric squat position, 3 × 2 min/2 min, 6 –10 Hz, 4–6mm, side-alternating vibration. Calisthenics group: relaxation, breathing retraining, calisthenics exercises.	WBV vs. Calisthenic: no significant difference between groups for the 6MWT (SMD = $0.54$ , 95% CI: $-0.23$ , $1.32$ ).
Salhi et al. (61)	62 patients with COPD WBV-group: $n = 31$ , 21 men, 10 women, mean age 58 years. Conventional resistance training (RT): $n = 31$ , 23 men, 8 women, mean age 63 years.	12 weeks, 3 times a week, same pulmonary rehabilitation program. WBV-group: FITVIBE, Gymna, Belgium, 8 upper and lower body exercises, 1–3 sets/exercise, 30–60s/ $n$ 27 Hz, 2 mm, vertical vibrations. RT: lower and upper body exercises, 3 × 10 repetitions	WBV vs. RT: no significant difference between groups for the 6MWT (SMD = $-0.24$ , 95% CI: $-0.79$ , $0.31$ )
Pleguezuelos et al. (57)	51 stable male patients with COPD Whole Body Vibration Training Group: $n = 26$ , mean age 68.4 years. Control group: $n = 25$ , mean age 71.3 years.	6 weeks, 3 sessions per week, regular prescribed medical treatment. WBVTG: Gymnauni phy. Nv. Pasweg 6a 3740 Bilzen, Belgium, squatting position, 6 × 30s/60s, 35Hz and 2mm, vertical vibrations. Control Group: general recommendations about physical activity and lifestyle.	WBV vs. control: no significant difference between groups for the 6MWT (SMD = $2.59$ , 95% CI $1.83$ , $3.35$ ).
<b>STROKE</b>			
Alp et al. (28)	21 post stroke patients WBV: ( $n = 10$ ), 10 men, 0 women, mean age $61.20 \pm 11.043$ years. Control group: ( $n = 11$ ) 9 men, 2 women, mean age $62.9 \pm 8$ years	4 weeks, 3 days a week, stretching and active range of motion exercises on the hemiplegic lower extremity for 15 min. WBV: Compex Winplate by Uniphy Elektromedizin GmbH and CoKG, tiptoes, 3 × 10s/3–20 s, 5min, 40 Hz, 4 mm, $n$ Control group: same exercises, no vibration.	Lack of data post WBV. The groups were statistically different at baseline for the 10MWT ( $p < 0.001$ ). The authors reported a significant improvement of the 10MWT in favor of the WBV group ( $p < 0.001$ ).

(Continued)

TABLE 1 | Continued

Article	Subjects	Interventions	Outcomes (only intergroup differences are presented)
Choi et al. (72)	30 individuals who presented with a gait deviation after a first stroke (>6 months). WBV-Treadmill Training (TT): $n = 15$ , 8 men, 7 women, mean age 51.93 years. TT group: $n = 15$ , 11 men, 4 women, mean age 53.67 years.	6 weeks, 3 times a week, 20 min of TT for both groups. WBV- TT: Galileo 2000, (Novotec, Germany, 2011), dynamic exercises, 6 × 45s/1min20-30 HZ, 3 mm, side-alternating vibration. TT group: same exercises on the platform without vibration.	No significant difference between group for the Walking speed (SMD = 0.32, 95% CI: -0.40, 1.04) and stride length (SMD = 0.50, 95% CI: -0.23, 1.23).
Choi et al. (38)	22 individuals who were diagnosed with strokes at least 6 months prior to the study. WBV group: $n = 11$ , 8 males, 3 females, mean age 50.9 years. Control group: $n = 11$ , 7 males, 4 females, mean age 52.2 years.	4 weeks, 5 times per week. WBV group: Galileo tilt table (Novotec Medical, Germany), squat posture, 10 min/session, 25 Hz, 5 mm, $n$ Control group: 30 min of Neuro-developmental treatment as the experimental group.	WBV vs. control group: no significant difference between group for the TUG test (SMD = -0.50, 95% CI : -1.35, 0.35).
Liao et al. (54)	84 individuals with hemispheric stroke persisting for more than 6 months before the time of enrolment low-intensity WBV group (LWBV): $n = 28$ , 20 men, 8 women, mean age 60.9 years High-intensity WBV group (HWBV): $n = 28$ , 18 men, 10 women, mean age 62.9 years Control (CON): $n = 28$ , 24 men, 4 women, mean age 59.8 years.	3 times a week, 30 sessions, same dynamic and static exercises, Gymna Fitvibe Medical System, Gymna Uniphy Pasweg, Bilzen, Belgium, synchronous vibrations. LWBV: 20 Hz, 1 mm HWBV: 30 Hz, 1 mm Control (CON): standing on the same WBV platform turned off.	LWBV vs. control: no significant difference between group for the 6MWT (SMD = 0.05 95% CI:-0.47, 0.58) and for TUG test (SMD = -0.10 95% CI: -0.62, 0.43). HWBV vs. Control: no significant difference between group for the 6MWT (SMD = -0.03, 95% CI-0.55, 0.50) and for TUG test (SMD = -0.25, 95% CI: -0.77, 0.28).
Lau et al. (51)	82 chronic stroke patients. WBV group: $n = 41$ , 26 men, 15 women, mean age 57.3 years. Control group: $n = 41$ , 32 men, 9 women, mean age 57.4 years.	8 weeks, 3 times a week. WBV: Jet-Vibe System (Danil SMC Co. Ltd., Seoul, South Korea), dynamic leg exercises, 6 exercises, 9-15 × 1,5-2, 5min/ $n$ , 20-30 Hz, 0.44-0.60 mm, vertical vibrations. Control group: same exercises without vibration.	WBV vs. control: no significant improvement of the 6MWT (SMD = -0.22 95% CI = -0.66, 0.21) and the 10MWT (SMD = 0.39 95% CI: -0.05, 0.83).
Brogårdh et al. (35)	31 individuals with chronic stroke. WBV: $n = 16$ , 13 men, 3 women, mean age 61.3 years. Control group: $n = 15$ , 12 men, 3 women, mean age 63.9 years.	6 weeks, 2 sessions/week. WBV training: Xrsize, static position knee flexed, 4-12 × 40-60s/1min, 25Hz, 3.75 mm, vertical vibrations. Control group: placebo vibrating platform (25 Hz, 0.2 mm amplitude).	WBV vs. control group: groups were different at baseline for the TUG test and the 6MWT. The authors reported significant improvements in both outcomes after WBV ( $p < 0.05$ )
van Nes Ilse et al. (67)	53 post-stroke patients WBV group: $n = 27$ , 16 males, 11 females, mean age of 59.7 years. Exercise therapy on music group: $n = 26$ , 14 males, 12 females, mean age of 62.6 years.	6 weeks, 5 days per week, physical therapy WBV group: Galileo 900, Galileo 2000, Enschede, The Netherlands, squat position hips and knees slightly flexed, 4 × 45 s/60 s, 30 Hz, 3 mm, Side-alternating vibration. Exercise therapy on music group: same standing position, exercises and relaxation.	WBV vs. exercise therapy on music group: no significant difference between groups (SMD = 0.00, 95% CI: -0.54, 0.54).
<b>KNEE OSTEOARTHRITIS</b>			
Bokaeian et al. (33)	28 patients with knee osteoarthritis. WBV + Strength Training (ST): $n = 15$ , 0 men, mean age 51.8 years Strength Training: $n = 13$ , 2 men, 11 women, mean age 54 years.	8 weeks, 3 times a week, both groups received same PT and strengthening exercises protocol. WBV + ST group: Fitvibe device (Italy), bent knees, 6-9 × 30-70 s/30-70 s, 25-30 HZ, 2 mm, progressively increased, vertical vibrations. Strength Training: flexion and extension exercise of knee joint, 3 sets, 10 RM, progressive load.	Lack of data post WBV. The authors reported significant improvements for the 2MWT, 50FWT and TUGT in favor of the WBV + ST group ( $p = 0.009$ ).

(Continued)

TABLE 1 | Continued

Article	Subjects	Interventions	Outcomes (only intergroup differences are presented)
Wang et al. (68)	49 patients with knee osteoarthritis. Whole Body Vibration Exercise + Quadriceps Resistance Exercise group: $n = 49$ , 13 men, 36 women, mean age 61.2 years. Quadriceps Resistance Exercise only group: $n = 50$ , 15 men, 35, women, mean age 61.5 years.	24 weeks, 5 days/week. WBV + QRE: My7TM model Personal Plate, Power Plate, USA, knees slightly flexed, 30 × 60 s/60 s, 35 Hz, 4–6 mm, $n$ Quadriceps Resistance Exercise: static and dynamic exercises.	WBV + QRE vs. QRT: significant improvement in favor of the WBV+QRE group for the TUG test (SMD = -3.11 95% CI: -3.71, -2.52) and the 6MWT (SMD = 1.68, 95% CI: 1.22, 2.14)
Simão et al. (63)	31 elderly subjects with knee osteoarthritis. WBV group: $n = 10$ , mean age 75 years. Squat group: $n = 10$ , mean age 73.4 years. Control group: $n = 11$ , mean age 71 years.	12 weeks, 3 times per week. WBV group: FitVibe, squat exercise, 6–8 × 20–40 s/20–40 s, 35–40 Hz, 4 mm, vertical vibrations. Squat group: 3s of isometric contraction. Control group: without intervention. No change their lifestyle.	WBV vs. Control group: no significant difference between groups for the 6MWT (SMD= 0.56, 95% CI: -0.26, 1.37) and gait speed (SMD= 0.39, 95% CI -0.42, 1.20).
Avelar et al. (29)	21 elderly patients with knee osteoarthritis. WBV group: $n = 11$ , mean age 75 years. Control group: $n = 10$ , mean age 71 years.	12 weeks, 3 times per week. WBV: FitVibe, GymnaUniphy NV, Bilzen, Belgium, squat training with WBV, 6–8 × 20–40 s/20–40 s, 35 Hz–40 Hz, 4 mm, $n$ Control: squat training without vibration.	WBV vs. control group: no significant difference between groups for the TUG test (SMD = 0.06, 95% CI: -0.80, 0.91). Significant improvement of the 6MWT in favor of the WBV group (SMD = 1.49, 95% CI: 0.49, 2.48).
<b>POSTMENOPAUSAL WOMEN</b>			
Sucuoglu et al. (66)	42 postmenopausal women patients WBV- Balance Coordination Exercise (BCE) group: $n = 21$ , mean age 56 years. BCE group: $n = 26$ , mean age 58.76 years.	4 weeks, 10 sessions per week, identical BCE programs. WBV-BCE group: Power Plate (Performance Health Systems UK Ltd, London, UK), 3 static positions, 2 × 30 s/60 s 30–35 Hz. 5 sessions per week, vertical vibrations. BCE group: 20-min exercise sessions at home, twice per day.	Significant difference at baseline between groups for the TUG test. The authors reported significant improvement compared with pretreatment values in both groups ( $p < 0.005$ ).
Iwamoto et al. (48)	52 ambulatory postmenopausal women with osteoporosis. WBV group: $n = 26$ , mean age 72.4 years. Control group: $n = 26$ , mean age 76 years.	6 months, $n$ WBV group: Galileo machine (G- 900; Novotec, Pforzheim, Germany), bent knees, 4 min, 20 Hz, $n$ , side to side alternating vibrations. Control group: no exercise.	Lack of data post WBV. The authors reported no significant difference between groups for the TUG test ( $p > 0.05$ ) but significant improvement in favor of the WBV group for the 10MWT ( $p < 0.05$ ).
Raimundo et al. (58)	27 postmenopausal women WBV: $n = 14$ , mean age 66 years Walk based program: $n = 13$ , mean age 66 years.	3 times a week, 8 months WBV: Galileo 154 2000, Novotec GmbH, Pforzheim, Germany, static knees flexed at 120°, 3–6 × 1 min/1 min, 12.6 Hz, 6 mm, increased each week, side-alternating oscillations. Walk based program: 2 × 25 min of walk, 70–75% HRmax	Lack of data post WBV. The authors reported a significant improvement of the 10MWT post WBV ( $p = 0.006$ ).
<b>MULTIPLE SCLEROSIS</b>			
Ebrahimi et al. (39)	34 multiple sclerosis patients with mild to moderate disability WBV group: $n = 17$ , 5 men, 12 women, mean age 37.06 years. Control group: $n = 17$ 4 men, 13 women, mean age 40.75 years.	10 weeks, 3 times a week. WBV group: $n$ , static positions, 15 × 30s–2 min/30 s–5 min, 2–20 Hz, 2 mm, $n$ Control group: continued their normal life	WBV vs. control group: no significant difference between groups for the TUG test (SMD = -0.47, 95% CI: -1.20, 0.26). Significant improvement in favor of the WBV group for the 10MWT (SMD = -1.05, 95% CI: -1.82, -0.28) and the 6MWT (SMD = 1.22, 95% CI: 0.43, 2.01).
Broekmans et al. (34)	25 ambulatory community- based patients with multiple sclerosis. WBV: $n = 11$ , 7 men, 4 women, mean age 46.1 years Control: $n = 14$ , 11 men, 3 women, mean age 49.7 years	5 sessions per 2-week cycle, 20 weeks WBV: Alpha Vibe Nijverdal, The Netherlands, 2–5 static and dynamic leg squats and lunges, 1–3 × 30–60 s/30–120 s, 20–45 Hz, 2.5 mm, increased progressively, vertical vibrations. Control: maintain their usual lifestyle	Groups were statistically different at baseline for the TUG test and 2MWT. The authors reported no significant effects in both groups for the TUG test ( $p = 0.26$ ) and the 2MWT ( $p = 0.25$ ).

(Continued)

TABLE 1 | Continued

Article	Subjects	Interventions	Outcomes (only intergroup differences are presented)
<b>OTHER PATHOLOGIES IN ADULTS</b>			
In et al. (47)	28 patients who were diagnosed with incomplete cervical spinal injury WBV group: $n = 14$ , 9 men, 5 women, mean age $46.1 \pm 9.8$ years Control group: $n = 14$ , 10 men, 4 women, mean age $49.9 \pm 9.3$ years	8 weeks, 5 days a week, twice a day, conventional physical therapy WBV group: TT2590X7, TurboSonic Co., South Korea, semi-squat with slight flexion ( $140^\circ$ ) at hips, knees and ankles, $4 \times 45$ s/1 min, 30Hz, 2–4 mm, vertical vibrations. Control group: 16 min of placebo WBV and 30 min of physical therapy.	WBV vs. Control group: No significant difference between groups for the TUG test (SMD = $-0.64$ , 95% CI: $-1.40$ , $0.13$ ) and the 10MWT (SMD = $-0.23$ , 95% CI: $-0.97$ , $0.52$ ).
Gerhardt et al. (42)	22 adult patients with stable, symptomatic pulmonary arterial hypertension (PAH). WBV group: $n = 11$ , 7 men, 4 women, mean age 65.1 years. Control group: $n = 11$ , 6 men, 5 women, 46 years.	4 weeks, 16 sessions. WBV group: Galileo MedM plat- form (Novotec Medical GmbH, Pforzheim, Germany), specific coordination exercises, $n$ , 20 Hz, 20 mm, side alternating vibrations. Control group: received WBV in a second phase.	Lack of data post WBV. The authors indicated that WBV was associated with a significant improvement of the 6MWD versus baseline of $+38.6 \pm 6.6$ m ( $p < 0.001$ )
Gloeckl et al. (43)	83 patients after lung transplantation. WBVT group: $n = 34$ , 16 men, 18 women, mean age 56 years. Control group: $n = 36$ , 22 men, 14 women, mean age 56 years.	4 weeks, 3 times per week, same pulmonary rehabilitation program. WBVT group: GALILEO, Novotec Medical GmbH, Pforzheim, Germany, squat exercises, $4 \times 2$ min/4min, 24–26 Hz, 6 mm, side alternating vibrations. Control group: same squat training on the floor.	lack of data post WBV The authors reported a between group difference of 28 m (95%CI: 3 m to 54 m, $p = 0.029$ ) significantly different in favor of WBVT.
Gaßner et al. (41)	17 participants diagnosed with idiopathic Parkinson's disease. WBV group: $n = 8$ , 6 men, 2 women, mean age 71.4 years. Placebo group: $n = 9$ , 7 men, 2 women, mean age 68.2 years.	5 weeks, 2–3 times a week. WBV group: SRT Zeptor Medical plus noise, static position, $5 \times 60$ s/60 s, knees slightly bents, 6Hz, 3 mm, $n$ Placebo group: stood on the vibration platform in the same basic position.	WBV vs. placebo group: no significant difference between group for the TUG test (SMD = $-0.37$ , 95% CI: $-1.34$ , $0.59$ ), velocity (SMD = $-0.21$ , 95% CI: $-1.17$ , $0.74$ ) and step length (SMD = $0.14$ , 95% CI: $-0.81$ , $1.09$ ).
Johnson et al. (49)	16 individuals, 3–6 weeks post total knee arthroplasty WBV: $n = 8$ , 6 men, 2 women, mean age 67 years Traditional Progressive Resistance Exercise (TPRE): $n = 8$ , 4 men, 4 women, mean age 68.5 years	3 session a week, 4 weeks WBV: Power Plate, Badhoevendorp, The Netherlands, static and dynamic exercises, 4–6 exercises/session, 1–3 set/exercise, 30–60 s/ $n$ , 35 Hz, 2–5 mm, $n$ TPRE: 1–3 SET of 10 REP for strengthening exercises for lower limbs, exercises were progressed once the patient could complete the exercise	WBV vs. TRPE: no significant difference for the TUG test (SMD = $-0.59$ , 95% CI: $-1.59$ , $0.42$ ).
Ahlborg et al. (27)	14 persons with cerebral palsy, spastic diplegia WBV: $n = 7$ , 4 men, 3 women, mean age 32 years Resistance training: $n = 7$ , 4 men, 3 women, mean age 30 years	Three times weekly, 8 weeks, same warming up and stretching WBV: NEMES-LSC (Nemesis BV, Hengelo, The Netherlands), standing position, hips and knees in $50^\circ$ of flexion, $1-4 \times 30-110$ s/15–120 s, increased progressively, 11 levels of intensity, 25–40 Hz, 7/10 on the Borg Scale, $n$ Resistance training: leg press, 3 SET of 10–15 REP, progressive load.	WBV vs. RT: no significant difference between groups for the TUG test (SMD = $0.28$ , 95% CI: $-0.77$ , $1.34$ ).
<b>OTHER PATHOLOGIES IN CHILDREN</b>			
Högler et al. (46)	24 children (5–16 years) with clinically mild to moderate osteogenesis imperfecta. WBV training: $n = 12$ , 6 men, 6 women, mean age 9.38 years. Control group: $n = 12$ , 6 men, 6 women, mean age 6.49 years.	5 months, twice-daily, home use WBV training: Galileo MTM, Novotec Medical, Pforzheim, Germany), static and dynamic exercises, $3 \times 3$ min/3 min, 20–25 Hz, side alternating vibrations. Control group: continued to receive regular care.	Lack of data post WBV. The authors reported no significant difference between groups for the 6MWT ( $p = 0.278$ )

(Continued)

TABLE 1 | Continued

Article	Subjects	Interventions	Outcomes (only intergroup differences are presented)
Cheng et al. (37)	16 children with cerebral palsy, 8 boys and 8 girls, mean age 9.2 years. WBV group: $n = 8$ , $n$ Control group: $n = 8$ , $n$	8-week WMV intervention followed by an 8-week control condition, with a 4-week rest (crossover study). WBV: AV-001A, Body Green, Taipei, Taiwan, static position, 10 min, 20 Hz, 2 mm, vertical vibrations. Control: same procedure with the machine turned off.	Lack of data post WBV. The authors reported a significant difference between the treatment and control condition for the 6MWT ( $p = 0.005$ ).
Lee and Chon. (52)	30 patients with either the spastic diplegia or quadriplegia forms of cerebral palsy WBV group: $n = 15$ , 6 men, 9 women, mean age 10 years. Control group: $n = 15$ , 9 men, 6 women, mean age 9.66.	8 weeks, 3 days per week, conventional PT. WBV: Galileo system (Novotec Medical GmbH, Pforzheim, Germany), squat position, $6 \times 3$ min/3 min, 5–25 Hz, 1–9 mm, side alternating vibrations. Control group: conventional physical therapy training.	WBV vs. control group: Significant improvement in favor of the WBV group for the gait speed (SMD = 1.41, 95% CI: 0.60, 2.22) and stride length (SMD = 0.91, 95% CI: 0.15, 1.67)
Ruck et al. (60)	20 children with cerebral palsy WBV: $n = 10$ , 8 boys, 2 girls mean age 8.3 years Control: $n = 10$ , 6 boys, 4 girls, mean age 8.1 years	Physiotherapy according to the established school program, 6 months, 5 days per week WBV: Vibraflex Home Edition II, Orthometrix Inc, White Plains, NY. Outside of North America, Galileo Basic, knees and hips flexed 10–45°, dynamic exercises, $3 \times 3$ min/3 min, 12–18 Hz, 2–6 mm, side to side alternating vertical vibrations. Control: Physiotherapy only	Lack of data post WBV. The authors reported a significant improvement of the 10MWT in favor of the WBV ( $p = 0.03$ ).

WBV, Whole body vibration; TUG, Timed up and go test; 6MWT, 6-minute walk test; 2MWT, 2-minute walk test; 50FWT, 50-feet walk test; 10MWT, 10-meter walk test.

score of  $7.2 \pm 1.7$ . The poorest methodological quality was found for postmenopausal women with a mean score of  $4.7 \pm 1.1$ .

### Studies Included for Meta-Analysis

A total of 25 studies were included in statistical analysis. Eleven studies were included for meta-analysis in the elderly (30–32, 40, 44, 45, 50, 53, 59, 64, 69), four studies for COPD patients (57, 61, 65, 71), four studies for stroke patients (35, 38, 51, 54), four studies for patients with knee OA (29, 33, 63, 68) and two studies for patients with Multiple Sclerosis (MS) (3, 34).

### Results Ranked According to Aging and Pathology Elderly Subjects

Sixteen studies examined the effect of WBV on elderly subjects (9, 30–32, 36, 40, 44, 45, 50, 53, 55, 56, 59, 62, 64, 69). The studies had an average PEDro score of  $5.5 \pm 1.0$ . The sample size ranged from 19 to 159 participants with a mean age of  $76.5 \pm 5.8$  years. Most of the studies included both men and women except for three with women only (32, 55, 62) and one with only men (45). Only one study failed to mention the eligibility criteria (44) and seven studies exhibited heterogeneity in their baselines (9, 36, 40, 44, 55, 62, 64). Training duration varied from 6 weeks to 8 months. Fifteen studies had a frequency of three sessions per week (9, 30–32, 36, 40, 44, 45, 50, 53, 55, 56, 59, 64, 69) while one study involved two sessions per week (62). The frequency and amplitude of platform vibrations varied from 10 to 40 Hz and 0.5 to 8 mm, respectively. Intensity

was progressively increased in 11 studies (9, 30, 32, 40, 45, 50, 53, 55, 59, 62, 64). Eight studies used synchronous vibrations (9, 30, 31, 36, 40, 50, 59, 69) while the other eight studies (32, 44, 45, 53, 55, 56, 62, 64) did not mention the type of vibrations delivered by their devices. The number of vibration bouts delivered per session varied from two to 39 sets with a period lasting between 15 and 3 min each. Resting time was between 5 s and 5 min. In nine protocols (9, 30, 31, 36, 40, 44, 53, 56, 69), the subjects maintained a static position, while they performed dynamic exercises in three studies (32, 50, 55), or both in four studies (45, 59, 62, 64). The most frequently used outcome was TUG, found in 14 studies (9, 30–32, 36, 40, 45, 50, 53, 55, 59, 62, 64, 69). Six studies combined TUG with the Tinetti gait score (9, 30, 31, 36, 40, 64). Four studies assessed gait speed using the 10MWT (32, 45, 59, 69). Three studies assessed functional performance with the 6MWT (44, 50, 56). Two studies used the Locometrix system for biomechanical analysis (31, 36).

### Comparisons to control groups

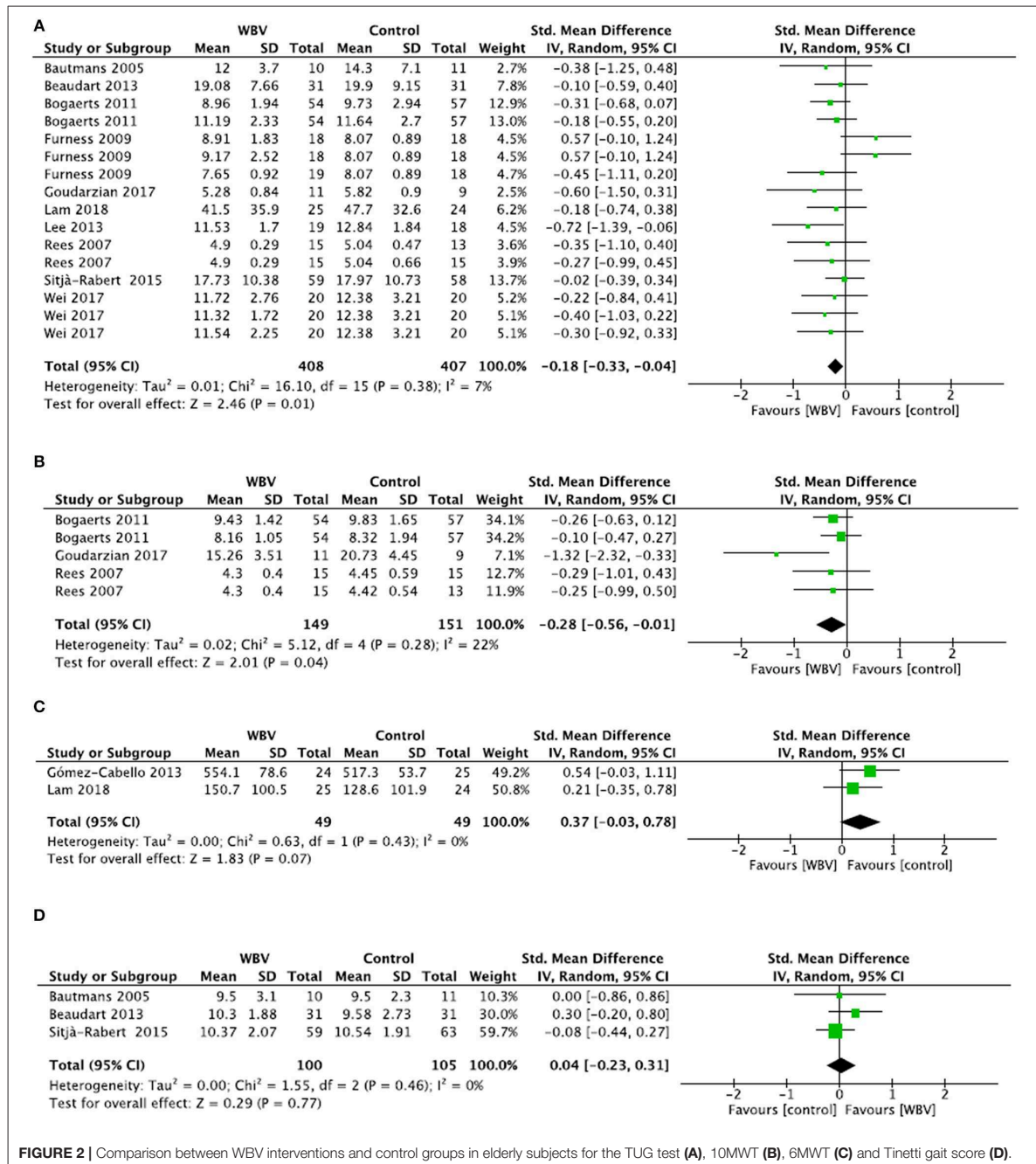
Four meta analyses (9, 31, 36, 55) were conducted for the following outcomes: TUG test, 10MWT, Tinetti test and 6MWT.

For the TUG test (Figure 2A), 10 studies were included in meta-analysis and four studies were excluded due to a lack of data despite requests to the authors (9, 31, 36, 55). Meta-analysis showed a significant decrease in time in favor of the WBV groups (SMD =  $-0.18$ ; 95% CI:  $-0.33, -0.04$ ), with consistent results ( $I^2 = 7\%$ ,  $p = 0.38$ ). The included studies were of high quality

**TABLE 2 |** Quality assessment with the PEDro scale.

Article	Items by number on the PEDro scale											Total score	Subjects
	1	2	3	4	5	6	7	8	9	10	11		
Lam et al. (50)	y	y	y	y	n	n	y	n	n	y	y	6	Older adults
Wei et al. (69)	y	y	n	y	n	n	n	y	y	y	y	6	
Goudarzian et al. (45)	y	y	n	y	n	n	n	y	n	y	y	5	
Sitjā-Rabert et al. (64)	y	y	y	n	n	n	y	n	y	y	y	6	
Santin-Medeiros et al. (62)	y	y	n	n	n	n	n	n	y	y	y	4	
Buckinx et al. (36)	y	y	n	n	n	n	y	n	y	y	y	5	
Lee et al. (53)	y	y	n	y	n	n	n	y	n	y	y	5	
Beaudart et al. (31)	y	y	n	y	n	n	n	y	y	y	y	7	
Gómez-Cabello et al. (44)	n	y	n	n	n	n	n	y	y	y	y	5	
Bogaerts, et al. (32)	y	y	y	y	n	n	n	y	n	y	y	6	
Machado et al. (55)	y	y	n	n	n	n	y	y	n	y	y	5	
Mikhael et al. (56)	y	y	y	y	y	n	y	n	n	y	y	7	
Furness and Maschette (40)	y	y	n	n	n	n	n	y	y	y	y	5	
Rees et al. (59)	y	y	n	y	n	n	n	y	n	y	y	5	
Bautmans et al. (30)	y	y	y	y	y	n	y	y	n	y	y	8	
Bruyere et al. (9)	y	y	n	n	n	n	n	y	y	y	y	5	
Spielmanns et al. (71)	y	y	y	y	n	n	n	n	n	y	y	5	COPD
Spielmanns et al. (65)	y	y	n	y	n	n	n	y	n	y	y	5	
Salhi et al. (61)	y	y	y	y	n	n	n	n	n	y	y	6	
Pleguezuelos et al. (57)	y	y	n	y	n	n	n	y	n	y	y	5	
Alp et al. (28)	y	y	y	n	n	n	y	y	n	y	y	6	Stroke
Choi et al. (72)	y	y	y	y	n	n	y	y	y	y	y	8	
Choi et al. (38)	y	y	n	y	n	n	n	n	n	y	y	4	
Liao et al. (54)	y	y	y	y	n	n	y	y	y	y	y	8	
Lau et al. (51)	y	y	y	y	n	n	y	y	y	y	y	8	
Brogårdh et al. (35)	y	y	y	y	y	n	y	y	y	y	y	9	
van Nes IJse et al. (67)	y	y	y	n	y	n	y	y	y	y	y	8	
Bokaeian et al. (33)	y	y	y	y	n	n	n	y	n	y	y	6	
Wang et al. (68)	y	y	y	y	n	n	y	y	y	y	y	8	
Simão et al. (63)	y	y	y	y	n	n	y	y	n	y	y	7	
Avelar et al. (29)	y	y	n	y	n	n	n	y	n	y	y	5	
Sucuoglu et al. (66)	y	y	n	y	n	n	n	n	n	y	y	4	Postmenopausal women
Iwamoto et al. (48)	y	y	n	y	n	n	n	y	y	y	y	6	
Raimundo et al. (58)	y	y	n	y	n	n	n	n	n	y	y	4	
Ebrahimi et al. (39)	y	y	n	y	n	n	n	y	n	y	y	5	Multiple sclerosis
Broekmans et al. (34)	y	y	y	y	n	n	y	y	n	y	y	7	
In et al. (47)	y	y	y	y	y	n	y	y	n	y	y	8	Other pathologies in adults
Gerhardt et al. (42)	y	y	n	y	n	n	n	n	n	y	y	4	
Gloeckl et al. (43)	y	y	y	y	n	n	y	n	n	y	y	6	
Gaßner et al. (41)	y	y	n	y	n	n	n	y	n	y	y	5	
Johnson et al. (49)	y	y	n	n	n	n	n	n	n	y	y	3	
Ahlborg et al. (27)	y	y	n	y	n	n	n	y	y	y	y	6	
Högler et al. (46)	y	y	y	y	n	n	n	n	n	y	y	5	Other pathologies in children
Cheng et al. (37)	y	y	n	y	n	n	n	y	y	y	y	6	
Lee et al. (53)	y	y	y	y	n	n	y	y	y	y	y	8	
Ruck et al. (60)	y	y	y	y	n	n	n	n	n	y	y	5	

*n*, criterion not fulfilled; *y*, criterion fulfilled; 1, eligibility criteria were specified; 2, subjects were randomly allocated to groups or to a treatment order; 3, allocation was concealed; 4, the groups were similar at baseline; 5, there was blinding of all subjects; 6, there was blinding of all therapists; 7, there was blinding of all assessors; 8, measures of at least one key outcome were obtained from more than 85% of the subjects who were initially allocated to groups; 9, intention-to-treat analysis was performed on all subjects who received the treatment or control condition as allocated; 10, the results of between-group statistical comparisons are reported for at least one key outcome; 11, the study provides both point measures and measures of variability for at least one key outcome; total score, each satisfied item (except the first) contributes 1 point to the total score, yielding a PEDro scale score that can range from 0 to 10. B, the level of evidence was B (randomized control trials that lacked double-blinding).



**FIGURE 2 |** Comparison between WBV interventions and control groups in elderly subjects for the TUG test (A), 10MWT (B), 6MWT (C) and Tinetti gait score (D).

(mean PEDro score = 5.8 ± 1.0), so a strong level of evidence supports the positive effect of WBV training on the TUG test.

For the 10MWT (Figure 2B), three studies were included in meta-analysis and one study was excluded because it used a different unit of measure (i.e., m/s instead of seconds in the other studies) (69). Meta-analysis showed a significant decrease

in time on the 10MWT in WBV groups (SMD = -0.28; 95% CI: -0.56, -0.01), with consistent results (I<sup>2</sup> = 22%, p = 0.28). The overall quality of the included studies was high (PEDro score = 5.0 ± 0.0). Thus, a strong level of evidence supports the positive effect of WBV training in improving gait speed on the 10MWT.

For the 6MWT (**Figure 2C**), two studies were included and one was excluded due to a lack of data despite requests to the authors (56). Meta-analysis showed no significant difference between groups (SMD = 0.37; 95% CI: -0.03, 0.78), despite a tendency toward an improvement in distance in WBV groups. Results were consistent ( $I^2 = 0\%$ ,  $p = 0.43$ ) and the quality of the included studies was high (PEDro score =  $5.5 \pm 0.7$ ). Thus, a strong level of evidence supports the lack of a beneficial effect of WBV training for improving performance in the 6MWT.

For the Tinetti gait score (**Figure 2D**), three studies were included in meta-analysis and three were excluded due to a lack of data despite requests to the authors (9, 31, 36). Meta-analysis showed no significant difference between groups (SMD = 0.04; 95% CI: -0.23, 0.31), with consistent results ( $I^2 = 0\%$ ,  $p = 0.46$ ). The quality of the included studies was high (mean PEDro score =  $7.0 \pm 1.0$ ). Thus, a strong level of evidence supports the absence of a positive effect of WBV training on the Tinetti gait score.

For biomechanical data recorded using the Locomotrix system (gait speed, stride frequency, stride length, stride symmetry, stride regularity, cranio-caudal mechanic power, antero-posterior mechanic power, medio-lateral mechanic power, and counting speed), no comparison between groups could be performed due to a lack of data despite requests to the authors (31, 36). Both Beaudart et al. (31) and Buckinx et al. (36) reported no significant inter-group difference for parameters recorded by the Locomotrix ( $p > 0.05$ ).

### Chronic COPD Patients

Four studies examined the effect of WBV on chronic COPD patients (57, 61, 65, 71) with an average PEDro score of  $5.2 \pm 0.5$ . The sample size ranged from 28 to 62 participants with a mean age of  $66.2 \pm 4.3$  years. Three studies included both men and women (61, 65, 71) and one included only male patients (57). All of the studies specified the eligibility criteria and had similar baselines. The training duration varied from 6 weeks to 3 months. In two studies (57, 61), subjects performed three WBV sessions per week, while patients had only two sessions per week in the other two studies (65, 71). The frequency and amplitude of the platform vibrations varied from 6 to 35 Hz and 2 to 6 mm, respectively. Intensity was progressively increased in two studies (65, 71). Half of the studies used side-alternating vibrations (65, 71) while the two other studies used synchronous vibrations (57, 61). The number of vibration bouts delivered per session varied from three to eight sets with a period lasting between 30 s and 2 min for each. Resting time was 60 s to 2 min. In two protocols (57, 71), the subjects maintained a static position, while they performed dynamic exercises in the other studies (61, 65). Only the 6MWT methodology was used to test gait.

### Comparisons to control groups (Figure 3A)

For the meta-analysis, two studies were included and two were excluded because the control groups were intervention groups with additional exercises not provided in the WBV group (i.e., not only WBV effects are measured) (61, 65). Meta-analysis showed no significant difference between groups (SMD = 1.66; 95% CI: -0.17, 3.49) with heterogeneous results ( $I^2 = 91\%$ ,

$p = 0.0008$ ). Thus, the level of evidence was conflicting for the 6MWT outcome in COPD.

For the excluded studies, Salhi et al. (61) showed that there was no significant difference between WBV training and conventional resistance training for improving 6MWT scores (SMD = -0.24; 95% CI: -0.79, 0.31). Similar results were found by Spielmanns et al. (65), where no significant difference was shown between the WBV intervention and the calisthenics group (SMD = 0.54; 95% CI: -0.23, 1.32).

### Comparison to pre-intervention (Figure 3B)

A second meta-analysis was conducted to include the four studies. Meta-analysis demonstrated a significant improvement in the distance walked during the 6MWT after WBV treatment (SMD = 0.92; 95% CI: 0.32, 1.51). Again, because there were heterogeneous results ( $I^2 = 66\%$ ,  $p = 0.03$ ), the level of evidence was conflicting for the 6MWT outcome.

### Stroke Patients

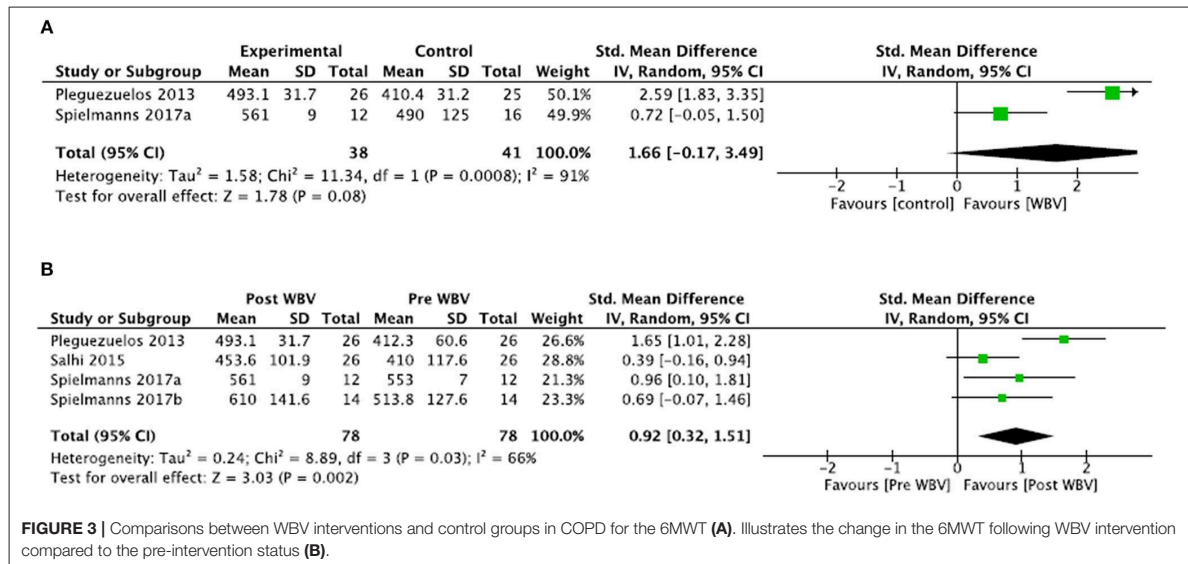
Seven studies examined the effect of WBV on stroke patients (28, 35, 38, 51, 54, 67, 72) with an average PEDro score of  $7.2 \pm 1.7$ . The sample size ranged from 21 to 84 participants with a mean age of  $58.3 \pm 4.5$  years. All of the studies included both men and women. All explained the eligibility criteria. Two studies (35, 38, 51, 72) found significant differences between groups for some outcomes at baseline. The training duration varied from 4 to 8 weeks. In four studies, subjects performed three sessions per week (28, 51, 54, 72), while patients had five sessions per week in two studies (38, 67), and two sessions per week in one study (35). The frequency and amplitude of platform vibrations varied from 20 to 40 Hz and 0.44 to 5 mm, respectively. The intensity was progressively increased in two studies (51, 72). Three studies used side-alternating vibrations (38, 67, 72), three synchronous vibrations (35, 51, 54), while one did not mention the type of vibrations (28). The number of vibration bouts delivered per session varied from 2 to 135 sets with a period lasting from 10 to 150 s each. Resting time was between 3 and 60 s. In four protocols (28, 35, 38, 67), the subjects maintained a static position, performed dynamic exercises in two studies (51, 72) and both types of exercises in one study (54). The TUG test was assessed in three studies (35, 38, 54), the 6MWT in three studies (35, 51, 54) and the 10MWT in two (28, 51). Only one study used a biomechanical methodology to assess gait function (72) and one study used the Functional Ambulation Categories (FAC) scale (67).

### Comparisons to control groups (Figures 4A and 4C)

Two meta-analyses were conducted for the TUG test and the 6MWT.

For the TUG test, two studies were included and one study was excluded because the groups were statistically different at baseline (35). Meta-analysis demonstrated no significant difference between groups (SMD = -0.21; 95% CI: -0.55, 0.13), with consistent results ( $I^2 = 0\%$ ,  $p = 0.83$ ). The quality of the study was high (mean PEDro score =  $8.0 \pm 0.0$ ). Thus, a strong level of evidence supports the absence of effect of WBV training on the TUG test in stroke patients.





For the 6MWT, two studies were included and one study was excluded because the groups were statistically different at baseline (35). Meta-analysis demonstrated no significant difference between the groups (SMD = -0.09; 95% CI: -0.37, 0.19), with consistent results ( $I^2 = 0\%$ ,  $p = 0.70$ ). The quality of the study was high (mean PEDro score =  $6.0 \pm 2.8$ ). Thus, a strong level of evidence supports the absence of effect of WBV training on the 6MWT test in stroke patients.

For biomechanical data, Choi et al. (72) demonstrated no significant difference between groups for stride length (SMD = 0.50; 95% CI: -0.23, 1.23) and walking speed (SMD = 0.32; 95% CI: -0.40, 1.04). Similarly, walking speed assessed by the 10MWT (51) was not different between groups (SMD = 0.39; 95% CI: -0.05, 0.83). Finally, the Functional Ambulation categories scale (67) was not different between groups (SMD = 0.00; 95% CI: -0.54, 0.54) after the interventions. All studies were of high quality RCT (Perdro scores  $\geq 5/10$ ). Thus, the level of evidence for each outcome was considered moderate.

#### Comparisons to pre-intervention (Figures 4B and 4D)

Two additional meta-analyses were conducted to include the two studies excluded for group comparisons for the TUG test and the 6MWT outcomes.

For the TUG test, meta-analysis showed a tendency but no significant improvement after the WBV treatment (SMD = -0.29; 95% CI: -0.60, 0.01) with consistent results ( $I^2 = 0\%$ ,  $p = 0.89$ ). The overall quality of the included studies was high (mean PEDro score =  $7.0 \pm 2.6$ ). Thus, a strong level of evidence supports the absence of effect of WBV treatment on the TUG test in stroke patients.

For the 6MWT, meta-analysis showed a significant improvement after WBV treatment (SMD = -0.33; 95% CI: 0.06, 0.59) with consistent results ( $I^2 = 0\%$ ,  $p = 0.58$ ). The overall quality of the included studies was high (mean PEDro

score =  $8.3 \pm 0.5$ ). Thus, a strong level of evidence supports the positive effect of WBV treatment to improve the distance walked during the 6MWT test in stroke patients.

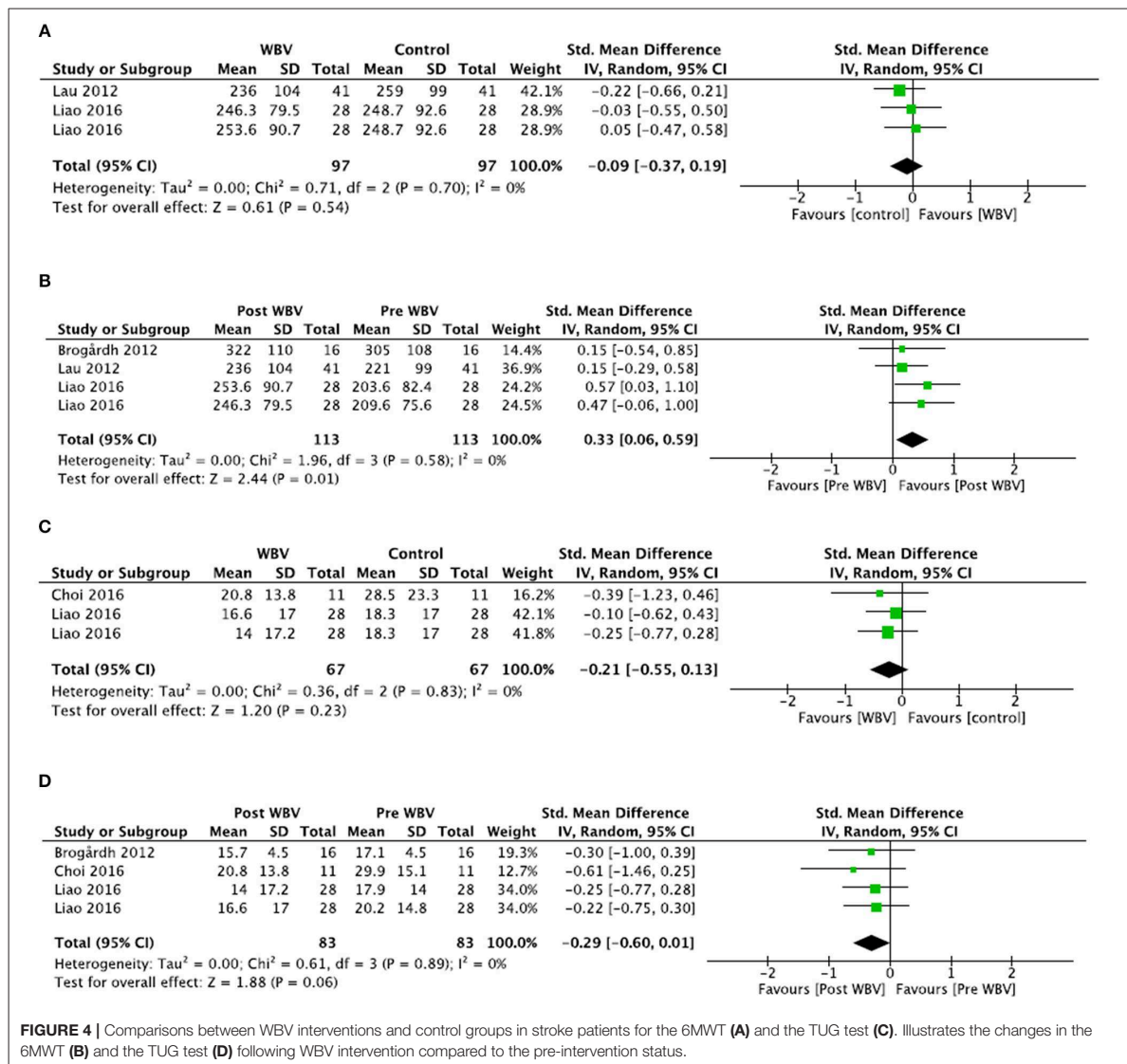
#### Knee Osteoarthritis

Four studies examined the effect of WBV on patients suffering from knee osteoarthritis (29, 33, 63, 68). The studies had an average PEDro score of  $6.5 \pm 1.2$ . The sample size ranged from 21 to 49 subjects with a mean age of  $65.1 \pm 9.2$  years. Two studies included both men and women (33, 68), while two studies did not mention the gender of the patients (29, 63). All of the studies specified the eligibility criteria and had similar baselines. The training duration ranged from 8 to 24 weeks. Three studies had a frequency of three sessions per week (29, 33, 63) while the other had five (68). The frequency and amplitude of the platform vibrations varied from 25 to 40 Hz and 2 to 6 mm, respectively. The intensity was progressively increased in all studies. Two studies used synchronous vibrations (33, 63) while two did not mention the type of vibrations of the devices (29, 68). The number of vibration bouts delivered per session varied from six to 30 sets with a period lasting 20 to 70 s. Resting time was between 20 and 70 seconds. In three protocols (29, 33, 68), the subjects maintained a static position, but performed static and dynamic exercises in the other study (63). Three studies used the TUG test (29, 33, 68), three used the 6MWT (29, 63, 68) and one combined the 2MWT and the 50FWT with the TUG (33).

#### Comparisons to control groups

For the TUG test (Figure 5B), two studies were included and one was excluded due to a lack of data (33). Meta-analysis showed no significant difference between groups (SMD = -1.54; 95% CI: -4.65, 1.56) with heterogeneous results ( $I^2 = 97\%$ ,  $p < 0.00001$ ). Thus, the level of evidence was conflicting.

For the 6MWT (Figure 5A), meta-analysis showed a significant difference in favor of the WBV group (SMD = 1.28;

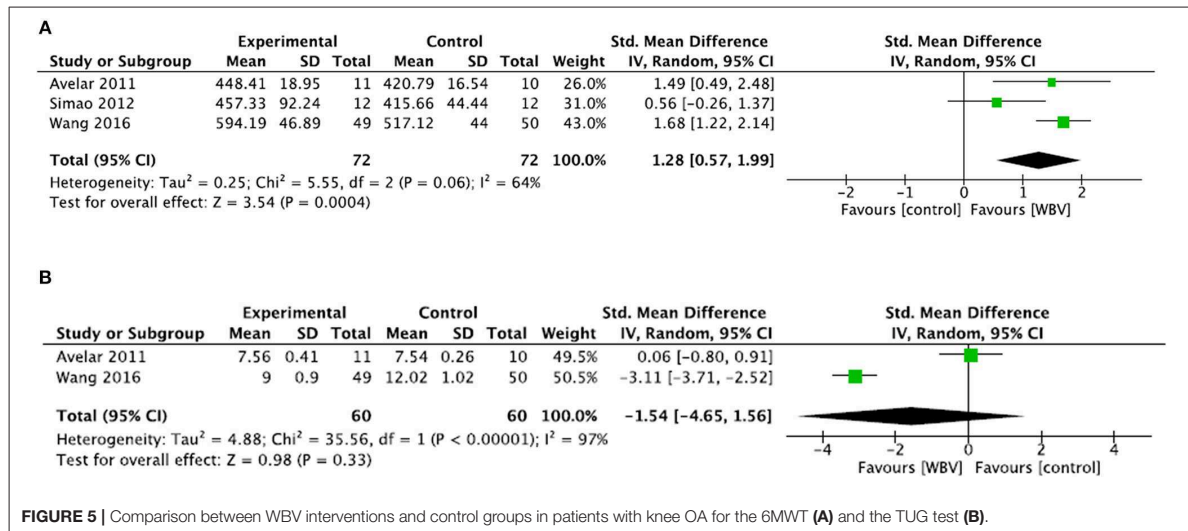


95% CI: 0.57, 1.99), with consistent results ( $I^2 = 64\%$ ,  $p = 0.06$ ). The quality of the studies was high (mean PEDro score =  $6.6 \pm 1.5$ ). Thus, a strong level of evidence supports the positive effect of adding WBV to improve the 6MWT in patients with knee OA.

### Postmenopausal Women

Three studies examined the effect of WBV on postmenopausal patients (48, 58, 66). These studies had an average PEDro score of  $4.6 \pm 1.1$ . The sample size ranged from 27 to 52 participants with a mean age of  $65.8 \pm 8.4$  years. All of the studies specified eligibility and had similar baselines. The training durations ranged from 4 weeks to 8 months. One study had a frequency of five sessions per week (66), another of three sessions per week

(58), while the last one did not specify the number of sessions per week (48). The frequency of the platform vibrations varied from 6 to 35 Hz and the amplitude was indicated in only one study (6 mm). The intensity of the sessions was progressively increased in two studies during training duration (58, 66). One study used synchronous vibrations (66) and the other two studies did not mention the type of vibrations (48, 58). The vibration bouts were delivered from 30 to 60 s with two to six sets. Resting time was 60 s in two studies (58, 66) and was not indicated in the third (48). In all protocols, the subjects maintained a static standing position. Two studies used the TUG (48, 66) and one combined it with a 10MWT (48). The third study measured walking speed along a four-meter pathway (58). Meta-analysis could not be



performed for the TUG test due to a lack of post-intervention data in all three studies, despite requests to the authors. Two studies reported significant improvement of the 10MWT after WBV training ( $p < 0.05$  and  $p = 0.006$ ) (48, 58). Sucuoglu et al. (66) showed a significant improvement of the TUG test post treatment ( $p < 0.005$ ), whereas Iwamoto et al. (48) found no significant difference between groups ( $p > 0.05$ ).

### Multiples Sclerosis

Two studies examined the effect of WBV on patients with multiple sclerosis (34, 39). The studies had an average PEDro score of  $6.0 \pm 1.4$ . The sample size ranged from 25 to 34 participants with a mean age of  $43.4 \pm 6.3$  years. Both studies included both men and women, specified the eligibility criteria and had similar baselines. The training duration was 10 and 20 weeks. In one study, patients underwent three sessions per week (39), while in the other they performed an average of 2.5 sessions per week (34). The frequency and amplitude of the platform vibrations varied from 2 to 45 Hz and 2 to 2.5 mm, respectively. The intensity was progressively increased in both studies. One study used synchronous vibrations (34) while the other did not mention the type of vibrations (39). The number of vibration bouts delivered per session varied from 2 to 15 sets with a period lasting 30 to 120 s. Resting time was between 30 and 120 s. In one protocol (39), the subjects maintained a static position, while they performed static and dynamic exercises in the other (34). Both studies used the TUG test. One study combined it with the 10MWT and the 6MWT (39), while the other used the TUG with the 2MWT and the 25-foot walk test (34).

### Comparisons to control groups

In one study, no meta-analysis was conducted for between-group comparisons because the groups were statistically different at baseline for the TUG test and 2MWT (34).

Ebrahimi et al. (39) found no significant difference between groups for the TUG test (SMD =  $-0.47$ ; 95% CI:  $-1.20, 0.26$ ).

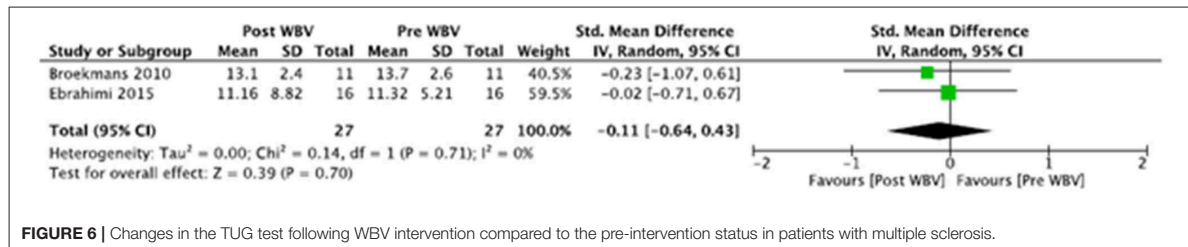
However, they did observe significant improvement in the WBV group for the 10MWT (SMD =  $-1.05$ ; 95% CI:  $-1.82, -0.28$ ) and the 6MWT (SMD =  $1.22$ ; 95% CI:  $0.43, 2.01$ ). The level of evidence was high (PEDro score = 5/10). Thus, the level of evidence was considered moderate for each outcome.

### Comparison to pre-intervention

Meta-analysis (Figure 6) showed no significant improvement in the TUG test after WBV training (SMD =  $-0.11$ ; 95% CI:  $-0.64, 0.43$ ) with consistent results ( $I^2 = 0\%$ ,  $p = 0.71$ ). The overall quality of the included studies was high (mean PEDro score =  $6.0 \pm 1.4$ ). Thus, there is a strong level of evidence to conclude that WBV treatment had no impact on the TUG test in patients with multiple sclerosis.

### Other Pathologies in Adults

Six studies reported results on different pathologies in adult patients (27, 41–43, 47, 49): incomplete cervical spinal injury (47), pulmonary arterial hypertension (42), lung transplantation (43), idiopathic Parkinson's disease (41), total knee arthroplasty (49) and cerebral palsy (27). The average PEDro score was  $5.3 \pm 1.7$ . The sample size ranged from 14 to 83 subjects with a mean age of  $54.6 \pm 14.1$  years. All of the studies included both men and women. All specified the eligibility criteria and had similar baselines except for one study where patients were statically different at baseline for certain outcomes (49). Training duration varied from 4 to 8 weeks. In four studies, subjects performed three WBV sessions per week (27, 41, 43, 49), while in one study they had four sessions per week (42), and five in another (47). The frequency and amplitude of the platform vibrations varied from 6 to 40 Hz and 2 to 20 mm, respectively. The frequency or amplitude of the vibrations was progressively increased in four studies (27, 43, 47, 49). Two studies used side-alternating vibrations (42, 43), one study used synchronous vibrations (47) and the other three studies did not mention the type of vibrations (27, 41, 49). The number of vibration bouts



**FIGURE 6 |** Changes in the TUG test following WBV intervention compared to the pre-intervention status in patients with multiple sclerosis.

delivered per session varied from 1 to 18 sets with a period lasting 30 to 120 s. Resting time was between 15 and 240 s. In three protocols (27, 41, 47), the subjects maintained a static position, while they performed dynamic exercises in two others (42, 43) and combined both in the last study (49). Four studies used the TUG test (27, 41, 47, 49) and three used the 6MWT (27, 42, 43) (one study combined both). One study used biomechanical analysis combined with the clinical TUG test (41). No meta-analysis was conducted due to the heterogeneity of patients in this subgroup.

#### Comparisons to control groups

In patients with incomplete cervical spinal injury, In et al. (47) found no significant difference between the WBV group and control group for the TUG test (SMD =  $-0.64$ ; 95% CI:  $-1.40, 0.13$ ) and the 10MWT (SMD =  $-0.23$ ; 95% CI:  $-0.97, 0.52$ ). The quality of the study was high (PEDro score = 8/10). Thus, the level of evidence was moderate.

In patients diagnosed with idiopathic Parkinson's disease, Gaßner et al. (41) observed no significant difference between the WBV group and the placebo group for the TUG test (SMD =  $-0.37$ ; 95% CI:  $-1.34, 0.59$ ), gait velocity (SMD =  $-0.21$ ; 95% CI:  $-1.17, 0.74$ ) and step length (SMD =  $0.14$ ; 95% CI:  $-0.81, 1.09$ ). The quality of the study was high (PEDro score = 5/10). Thus, the level of evidence was moderate.

After total knee arthroplasty, Johnson et al. (49) reported no significant difference for the TUG test between a WBV group and a resistance training group (SMD =  $-0.59$ ; 95% CI:  $-1.59, 0.42$ ). The quality of the study was low (PEDro score=3/10). Thus, the level of evidence was limited.

In patients with cerebral palsy, Ahlberg et al. (27) found no significant difference between a WBV group and a resistance training group for the TUG test (SMD =  $0.28$ ; 95% CI:  $-0.77, 1.34$ ). The quality of the study was high (PEDro score=6/10). Thus, the level of evidence was moderate.

SMD could not be reported in the study of Gerhardt et al. (42) and Gloeckl et al. (43) due to the lack of post-intervention data despite a request to the authors. The authors of the first study indicated that WBV was associated with a significant improvement of the 6MWD vs. baseline of  $+38.6 \pm 6.6$  m ( $p < 0.001$ ) (42). The authors of the second study reported a significant between-group difference of 28 m (95% CI: 3, 54;  $p = 0.029$ ) in favor of WBV (73).

#### Comparisons to pre-intervention

Ahlberg et al. (27) reported no significant difference after WBV (SMD =  $0.14$ ; 95% CI:  $-0.91, 1.19$ ) in patients with cerebral

palsy, with a level of evidence considered moderate (PEDro score = 6/10). Similarly, Johnson et al. (49) observed no significant improvement of the TUG test after WBV (SMD =  $1.02$ ; 95% CI:  $-0.04, 2.09$ ) in patients with total knee arthroplasty, with a limited level of evidence (PEDro score = 3/10).

#### Other Pathologies in Children

Four studies examined the effect of WBV in children(37, 46, 52, 60): two evaluated the effects of WBV in cerebral palsy (37, 60), one in patients with osteogenesis imperfecta (46) and one in patients with spastic diplegia or quadriplegia forms of cerebral palsy (52). The studies had an average PEDro score of  $6.0 \pm 1.4$ . The sample size ranged from 16 to 30 participants with a mean age of  $8.7 \pm 0.8$  years. All of the studies included both boys and girls, specified the eligibility criteria and had similar baselines. The training duration varied between 8 and 24 weeks. One study had a frequency of two sessions per week (46), one had three sessions per week (52), one had five sessions per week (60), while the number of sessions was not specified in the last study (37). The frequency and amplitude of the platform vibrations varied from 5 to 25 Hz and 1 to 9 mm, respectively. The frequency or amplitude of the vibrations was progressively increased in three studies (46, 52, 60). Two studies used synchronous vibrations (37, 60) and the other two studies used side-alternating vibrations (46, 52). The number of vibration bouts delivered per session varied from three to six sets with periods lasting 3 min each. Resting time was also 3 min. In two protocols (37, 52), the subjects maintained a static position, while they performed static and dynamic exercises in the other studies(46, 60). Two studies used the 6MWT (37, 46), and one study combined it with the TUG test (37). One study used biomechanical analysis including gait speed, stride length and cycle time (52). One study used the 10MWT (for gait speed) (60). Three studies reported significant improvement in gait parameters following WBV treatment (37, 52, 60), whereas one study reported that the 6MWT remained unchanged (46). No meta-analysis was conducted due to the heterogeneity of patients in this subgroup. Additionally, data post interventions were not reported in three studies (37, 46, 60).

#### Comparisons to control groups

In children with clinically mild to moderate osteogenesis imperfecta, Höglér et al. reported no significant difference between groups for the 6MWT ( $p = 0.278$ ) (46).

In children with cerebral palsy, Cheng et al. reported a significant difference between the treatment and control conditions for the 6MWT ( $p = 0.005$ ) (37). Ruck et al. reported

a significant improvement in the 10MWT in favor of WBV ( $p = 0.03$ ) (60).

Finally, in patients with either spastic diplegia or quadriplegia forms of cerebral palsy, Lee and Chon found significant improvement in favor of the WBV group for gait speed (SMD = 1.41; 95% CI: 0.60, 2.22) and stride length (SMD = 0.91; 95% CI: 0.15, 1.67) (52), with a level of evidence considered moderate (PEDro score = 8/10).

## DISCUSSION

The aim of this systematic review was to determine the changes in gait outcomes after WBV training in healthy adults and various patient categories. We found a strong level of evidence for a positive effect of WBV training on the TUG test and the 10MWT in the elderly. The same level of evidence was found in favor of a significant improvement of the 6MWT in stroke patients and patients with knee OA. In contrast, there is no change in the 6MWT and the Tinetti gait score in the elderly, and the TUG test was not improved in stroke or multiple sclerosis patients. Conflicting results were found in COPD patients despite significant improvements in the 6MWT. Other outcomes showed a moderate or limited level of evidence, due to a lack of data or because only one RCT was identified.

As mentioned in a prior review (20), the major obstacle in conducting meta-analysis and establishing strong evidence on the effects of WBV on gait is the heterogeneity in study methodologies. Intervention regimes, settings, combined interventions and control groups varied greatly. As in any training protocol, numerous factors can affect the results of the program (e.g., the duration of the intervention; the frequency or volume of the sessions; the type, frequency and amplitude of the vibrations and the exercises performed on the platform). Because the studies used different protocols, a random-effects model was used. In the presence of heterogeneity, a random-effects meta-analysis weights the studies relatively more equally than a fixed-effect analysis (25). Because control groups varied a great deal in terms of interventions (i.e., exercises, physical therapy, sham, no interventions etc.), intergroup comparisons were not always possible, so we added within group comparisons (i.e., pre vs. post WB) to estimate the effect of the treatment. Additionally, some groups were statistically different at baseline for certain outcomes (35, 40, 62), making between groups comparisons impossible after intervention. The results of each patient category are discussed below.

On the one hand, the results in the elderly showed significant improvements in the TUG test and the 10MWT after WBV intervention. These results are in favor of better dynamic stability and gait performance as both outcomes are related to balance and gait speed, respectively (45, 59).

However, the effect sizes were small ( $-0.18$  and  $-0.28$ , respectively) and the 6MWT was not significantly modified by treatment despite a tendency toward improvement in favor of WBV (SMD = 0.37; 95% CI:  $-0.03$ , 0.78). Our findings on physical improvements partially corroborate earlier reports. In a recent scoping review, Park et al. (74) concluded that WBV

training could be effective in increasing lean mass, muscular strength and cardiovascular health (74). Positive changes in body composition and fitness induced by WBV training may explain the improvement in gait performance. After the age of 50, muscle mass decreases approximately 2% every year and muscle strength decreases 15% every 10 years (75). These age-related changes impact functional mobility, including gait speed, static dynamic balance, and the risk of falling. As a resistance training exercise, WBV appears efficient in attenuating the loss of muscle mass and muscle strength. In order to combat the effects of aging, it should be recommended that older adults perform WBV 2 or 3 days per week, as suggested for resistance training (76, 77). In addition, because both offer numerous benefits, these interventions could be combined, depending on feasibility and patient motivation. These recommendations are also valuable for patients with reduced mobility and who need to improve their autonomy at home, as only the TUG test and 10MWT (short-distance walking tests) were improved but not the 6MWT (a long-distance walking test).

On the other hand, the results support that gait performance can be improved with no improvement in qualitative aspects of the locomotion pattern. In fact, the quality of gait, assessed by the Tinetti test, was not changed. The outcome is classically divided in two parts. One assesses static balance, while the second assesses dynamic balance (78, 89). Because gait was the main outcome of the present research, we did not include the total score in meta-analysis. All 28 points are necessary to assess the whole balance score, and thus, the risk of falling. Additionally, the Tinetti total score has been recently demonstrated as being related to muscle mass and strength (79). As previously discussed, improvements in muscle mass, strength, and performance are demonstrated after WBV training. Thus, changes in the Tinetti total score can be expected and further investigation on this outcome is warranted.

The results in stroke patients are more mitigated. On the one hand, between-group comparisons showed that the 6MWT was not modified by WBV training. However, two studies had to be excluded from this analysis and a second analysis was performed to include them in a pre vs. post comparison. This time, the results showed a significant increase in distance after a WBV intervention. On the other hand, no significant changes were found for the TUG test in both comparisons, despite a tendency toward an improvement. The 6MWT is commonly used to assess aspects of walking performance in stroke survivor studies (80). It evaluates the global responses of all the systems involved during exercise, including the pulmonary and cardiovascular systems, systemic circulation, peripheral circulation, blood, neuromuscular units, and muscle metabolism (90). The results concerning the 6MWT confirm the findings discussed previously in favor of functional improvements in elderly and disease populations (74). Conflicting results emerged from the TUG test. Balance, assessed by the TUG test, appeared less modified by the WBV treatment in the treatment of neuropathologic subjects than in the elderly population. This was confirmed by the poor results in multiple sclerosis patients included in meta-analysis for the TUG outcome (34, 39).

Conflicting results were also found for the 6MWT in COPD patients. The pooled studies demonstrated a significant improvement of the distance walked but with inconsistent results. The size of the overall effect was large (SMD = 0.92; 95% CI: 0.32, 1.51) and seems to corroborate the functional improvement observed after WBV. However, the results must be taken with caution because of their heterogeneity and the small sample of high quality studies revealed by the present review (i.e., four studies) (57, 61, 65, 71). Heterogeneity was found in both intergroup comparison and comparison with pre intervention, and was difficult to explain as the studies had similar populations (i.e., older adults with COPD) and settings (i.e., frequency and amplitude of the vibrations).

In patients with knee OA, a strong level of evidence supports the beneficial effect of WBV training in improving the 6MWT, with a large effect size (SMD=1.28; 95% CI: 0.57, 1.99). Interestingly, Wang et al. showed that adding WBV to quadriceps resistance training was more efficient than resistance training alone (SMD = 1.68; 95% CI: 1.22, 2.14) (68). A recent review showed that, in patients with knee OA, a resistance training program is effective for improving knee extensor strength but has limited effect on pain and disability if the gains are <30% (81). WBV interventions combined with strength training may help achieve this gain necessary for beneficial effects on pain and functional performance. Again, for the TUG test, Wang et al. (68) demonstrated the value of combining WBV and resistance training to improve gait performance (SMD = -3.11; 95% CI: -3.71, -2.52). However, because of the absence of effect in the second study (29), the level of evidence was conflicting. Heterogeneity might be explained by the two major differences between the studies, which were the addition of quadriceps resistance training and the doubled duration of the training (i.e., 24 vs. 12 weeks) in Wang et al. (68).

Although some significant gait improvements after WBV were reported, it is important to stress that significant statistical changes are not always linked to significant clinical improvements for patients. For example, in elderly patients, Bogaerts et al. (32) reported a significant improvement of the time required to perform the TUG test after WBV intervention. While the time was decreased from 13.1 s to 11.19 s (SMD = -0.71; 95%CI: -1.10, -0.32) with an effect size considered moderate, this difference may not correspond to major changes in the patients' daily activities. Thus, the benefit of a non-functional intervention such as WBV should always be questioned with regard to each patient's goals.

However, WBV appears to be a time-efficient and easy-to-use intervention that is both relatively inexpensive and safe for patients with balance deficits. Vibration plates are readily available at all rehabilitation hospitals/centers. Moreover, it might be interesting to complete certain conventional rehabilitation programs like resistance or balance training with WBV training that may offer the same results. For example, for COPD patients, it has been found that both the WBV and resistance training groups significantly improved in the 6MWT (61, 65, 71) with no significant difference between the groups. Additionally, no control group including any form of training was significantly superior to WBV training in improving gait.

Most of the studies included in this review (27/43 articles) reported a drop-out rate of <15% during their interventions. Considering that most of the subjects were patients with diseases or physical disorders, it is logical to assume that they would have stopped treatment had they experienced any harmful or adverse side effect. This might support the hypothesis that the patients tolerated vibration training well. Moreover, WBV training has been reported to be appreciated and considered a safe training method. The fact that participants could perform either dynamic or static exercises while holding a bar increases safety and would be beneficial for the weakest populations such as elderly persons with balance impairments.

The vibration type may impact the training response. side-alternating WBV has been shown to increase heart rate higher than synchronous vibrations in young sedentary women during 20-min sessions (82). This illustrates the potential of WBV to improve fitness capacity, particularly in less active populations. Additionally, higher electromyography of knee extensor and plantar flexor activities were observed with a side-alternating vibration platform compared to synchronous vibrations (83). Although these results are in favor of side-alternating vibrations, our review showed heterogeneous results regarding the vibration type when it was mentioned. Significant improvements in gait parameters were found in 13 studies that used synchronous vibrations (9, 30, 33, 37, 40, 47, 51, 57, 59–61, 63, 66) and in seven studies that used side-alternating vibrations (38, 42, 43, 48, 52, 65, 72). Again, because of the lack of consistency in the protocols and results, it is difficult to reach a consensus on specific WBV training to improve human locomotion.

We chose to select only studies on long-term effects because they are better correlated to conventional physiotherapies that often last many weeks. Moreover, long-term effects have been studied more than short-term or even immediate effects. We found that a wide range of protocols lasted 6 weeks or more and a few lasted 4 weeks. However, we can add that only a few RCT focused on the acute effects of WBV training on gait parameters (84–86). In the future, it might be interesting to compare different WBV protocols (i.e., with different WBV frequencies) in order to evaluate the effect of high vs. low WBV frequency on balance and gait within a single session.

Finally, most studies used clinical assessments instead of biomechanical analysis (41, 52, 72). Since it might be a more objective measure, future studies should integrate this kind of outcome more often in order to compare it with functional assessments.

## LIMITATIONS OF THE STUDY

Results of meta-analysis must be taken with caution as some studies could not be included in the comparisons due to a lack of complete data, notably for the TUG test and Tinetti gait score in the elderly, despite requests to the authors.

The Cochrane Qualitative and Implementation Methods Group recommends the application of Grades of Recommendation, Assessment, Development, and Evaluation (GRADE) in the Evidence from Qualitative Reviews to assess

confidence in qualitative synthesized findings (87). However, the GRADE necessitates assessing the risk of publication bias with a funnel plot, determining its asymmetry, which can be performed with at least ten studies (88). Because most of the statistical analyses were conducted on few RCT, we decided to implement other guidelines described by a Cochrane collaboration group to assess the level of evidence (26). Because this method includes fewer criteria, our confidence in the results must be taken with caution.

## CONCLUSION

While WBV training appears to be a useful and relatively successful tool in improving gait and walking abilities, it remains unclear whether the treatment could be generalized to all patients. Some populations have been studied more than others with varying degrees of consistency. In the elderly, there is a strong level of evidence that WBV can improve mobility by improving the TUG test, and gait speed by improving the 10MWT. The results also showed significant improvements to functional performance in stroke patients and patients with knee OA by improving the 6MWT. However, the treatment

was inefficient in changing the TUG test in stroke and multiple sclerosis patients, and conflicting results were obtained for the 6MWT in COPD. Finally, other outcomes were studied less and the level of evidence was moderate or even limited depending of the quality of the study. The transferability of this kind of training to daily activities remains unclear and the use of vibration training to replace functional rehabilitation must always be questioned. Further research is needed to explore the possibility of finding a standardized protocol targeting gait ability in a wide range of populations.

## AUTHOR CONTRIBUTIONS

MF, TV, GL, PF, TH, LC, J-LH, EY, P-AD, and AD designed the study, collected, analyzed, and interpreted the data, drafted and revised the manuscript, tables and figures, and gave final approval.

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## REFERENCES

- Goetz CG. Jean-martin charcot and his vibratory chair for Parkinson disease. *Neurology*. (2009) 73:475–8. doi: 10.1212/WNL.0b013e3181b1640b
- Kaeding TS. Sarkopenie und vibrationstraining: eine übersicht. *Zeitschrift für Gerontologie Geriatrie*. (2009) 42:88–92. doi: 10.1007/s00391-008-0565-4
- Weber-Rajek M, Mieszkowski J, Niespodzinski B, Ciechanowska K. Whole-body vibration exercise in postmenopausal osteoporosis. *Prz Menopauzalny*. (2015) 14:41–7. doi: 10.5114/pm.2015.48679
- Wang XQ, Pi YL, Chen PJ, Chen BL, Liang LC, Li X, et al. Whole body vibration exercise for chronic low back pain: study protocol for a single-blind randomized controlled trial. *Trials*. (2014) 15:104. doi: 10.1186/1745-6215-15-104
- Collado-Mateo D, Adsuar JC, Olivares PR, del Pozo-Cruz B, Parraca JA, del Pozo-Cruz J, et al. Effects of whole-body vibration therapy in patients with fibromyalgia: a systematic literature review. *Evid Based Compl Alt Med*. (2015) 2015:719082. doi: 10.1155/2015/719082
- Cochrane DJ. The potential neural mechanisms of acute indirect vibration. *J Sports Sci Med*. (2011) 10:19–30.
- Aminian-Far A, Hadian MR, Olyaei G, Talebian S, Bakhtiary AH. Whole-body vibration and the prevention and treatment of delayed-onset muscle soreness. *Journal of Athletic Training*. (2011) 46, 43–49. doi: 10.4085/1062-6050-46.1.43
- Melnyk M, Kofler B, Faist M, Hodapp M, Collhofer A. Effect of a whole-body vibration session on knee stability. *Int J Sports Med*. (2008) 29:839–44. doi: 10.1055/s-2008-1038405
- Bruyere O, Wuidart M-A, Palma ED, Gourlay M, Ethgen O, Richey F, et al. Controlled whole body vibration to decrease fall risk and improve health-related quality of life of nursing home residents. *Arch Phys Med Rehabil*. (2005) 86:303–7. doi: 10.1016/j.apmr.2004.05.019
- Zaidell LN, Mileva KN, Summers DP, Bowtell JL. Experimental evidence of the tonic vibration reflex during whole-body vibration of the loaded and unloaded leg. *PLoS ONE*. (2013) 8. doi: 10.1371/journal.pone.0085247
- Martin BJ, Park HS. Analysis of the tonic vibration reflex: influence of vibration variables on motor unit synchronization and fatigue. *Eur J Appl Physiol Occup Physiol*. (1997) 75:504–11. doi: 10.1007/s004210050196
- Mileva KN, Bowtell JL, Kossev AR. Effects of low-frequency whole-body vibration on motor-evoked potentials in healthy men. *Exp Physiol*. (2009) 94:103–16. doi: 10.1113/expphysiol.2008.042689
- Games KE, Sefton JM, Wilson AE. Whole-body vibration and blood flow and muscle oxygenation: a meta-analysis. *J Athl Train*. (2015) 50:542–9. doi: 10.4085/1062-6050-50.2.09
- Rogan S, Taeymans J, Radlinger L, Naeflin S, Ruppen S, Bruelhart Y, et al. Effects of whole-body vibration on postural control in elderly: an update of a systematic review and meta analysis. *Arch. Gerontol. Geriatr*. (2017) 73:95–112. doi: 10.1016/j.archger.2017.07.022
- Takakusaki K. Functional neuroanatomy for posture and gait control. *J Mov Disord*. (2017) 10:1–17. doi: 10.14802/jmd.16062
- Breniere Y, Do MC. When and how does steady state gait movement induced from upright posture begin? *J Biomech*. (1986) 19:1035–40.
- Breniere Y, Do MC. Control of gait initiation. *J Mot Behav*. (1991) 23:235–40. doi: 10.1080/00222895.1991.9942034
- Winter D. Human balance and posture control during standing and walking. *Gait Posture*. (1995) 3:193–214. doi: 10.1016/0966-6362(96)82849-9
- Hendrickson J, Patterson KK, Inness EL, McIlroy WE, Mansfield A. Relationship between asymmetry of quiet standing balance control and walking post-stroke. *Gait Posture*. (2014) 39:177–81. doi: 10.1016/j.gaitpost.2013.06.022
- Lindberg J, Carlsson J. The effects of whole-body vibration training on gait and walking ability - a systematic review comparing two quality indexes. *Physiother Theory Pract*. (2012) 28:1–14. doi: 10.3109/09593985.2011.641670
- Natalie A de Morton. The PEDro scale is a valid measure of the methodological quality of clinical trials: a demographic study. *Aust J Physiother*. (2009) 55, 129–133. doi: 10.1016/S0004-9514(09)70043-1
- Wan X, Wang W, Liu J, Tong T. Estimating the sample mean and standard deviation from the sample size, median, range and/or interquartile range. *BMC Med Res Methodol*. (2014) 14. doi: 10.1186/1471-2288-14-135
- RevMan 5 download. (2019). Available online at: /help/tools-and-software/revman-5/revman-5-download (accessed March, 26 2019).
- Cohen J. *Statistical Power Analysis for the Behavioral Sciences, 2nd ed.* (1988). Hillsdale, N.J: L. Erlbaum Associates.
- Higgins J, Green, S, editors. *Cochrane Handbook for Systematic Reviews of Interventions*. Version 5.1.0. (2011) The Cochrane Collaboration. Available online at: www.cochrane-handbook.org (accessed March 2011).

26. van Tulder M, Furlan A, Bombardier C, Bouter L. Updated method guidelines for systematic reviews in the cochrane collaboration back review group. *Spine*. (2003) 28, 1290–1299. doi: 10.1097/01.BRS.0000065484.95996.AF
27. Ahlborg L, Andersson C, Julin P. Whole-body vibration training compared with resistance training: effect on spasticity, muscle strength and motor performance in adults with cerebral palsy. *J Rehabil Med*. (2006) 38:302–8. doi: 10.1080/16501970600680262
28. Alp A, Efe B, Adali M, Bilgiç, A, Demir Türe S, Coşkun S, et al. The impact of whole body vibration therapy on spasticity and disability of the patients with poststroke hemiplegia. *Rehabilit. Res. Prac*. (2018) 2018, 1–6. doi: 10.1155/2018/8637573
29. Avelar NCP, Simão AP, Tossige-Gomes R, Neves CDC, Rocha-Vieira E, Coimbra CC, et al. The effect of adding whole-body vibration to squat training on the functional performance and self-report of disease status in elderly patients with knee osteoarthritis: a randomized, controlled clinical study. *J Alt Compl Med*. (2011) 17:1149–55. doi: 10.1089/acm.2010.0782
30. Bautmans I, Van Hees E, Lemper JC, Mets T. The feasibility of whole body vibration in institutionalised elderly persons and its influence on muscle performance, balance and mobility: a randomised controlled trial [ISRCTN62535013]. *BMC Geriatr*. (2005) 5. doi: 10.1186/1471-2318-5-17
31. Beaudart C, Maquet D, Mannarino M, Buckinx F, Demonceau M, Crielaard JM, et al. Effects of 3 months of short sessions of controlled whole body vibrations on the risk of falls among nursing home residents. *BMC Geriatr*. (2013) 13:42. doi: 10.1186/1471-2318-13-42
32. Bogaerts A, Delecluse C, Boonen S, Claessens AL, Milisen K, Verschueren SMP. Changes in balance, functional performance and fall risk following whole body vibration training and vitamin D supplementation in institutionalized elderly women. A 6 month randomized controlled trial. *Gait Posture*. (2011) 33:466–472. doi: 10.1016/j.gaitpost.2010.12.027
33. Bokaeian HR, Bakhtiari AH, Mirmohammadhani M, Moghimi J. The effect of adding whole body vibration training to strengthening training in the treatment of knee osteoarthritis: a randomized clinical trial. *J Bodyw Movement Therap*. (2016) 20:334–40. doi: 10.1016/j.jbmt.2015.08.005
34. Broekmans T, Roelants M, Alders G, Feys P, Thijs H, Eijnde B. Exploring the effects of a 20-week whole-body vibration training programme on leg muscle performance and function in persons with multiple sclerosis. *J Rehabil Med*. (2010) 42:866–72. doi: 10.2340/16501977-0609
35. Brogårdh C, Flansbjerg U-B, Lexell J. No specific effect of whole-body vibration training in chronic stroke: a double-blind randomized controlled study. *Arch Phy Med Rehabil*. (2012) 93:253–8. doi: 10.1016/j.apmr.2011.09.005
36. Buckinx F, Beaudart C, Maquet D, Demonceau M, Crielaard JM, Reginster JY, et al. Evaluation of the impact of 6-month training by whole body vibration on the risk of falls among nursing home residents, observed over a 12-month period: a single blind, randomized controlled trial. *Aging Clin Exp Res*. (2014) 26:369–76. doi: 10.1007/s40520-014-0197-z
37. Cheng HYK, Yu YC, Wong AMK, Tsai YS, Ju YY. Effects of an eight-week whole body vibration on lower extremity muscle tone and function in children with cerebral palsy. *Res Dev Disabil*. (2015) 38:256–61. doi: 10.1016/j.ridd.2014.12.017
38. Choi E-T, Kim Y-N, Cho W-S, Lee D-K. The effects of visual control whole body vibration exercise on balance and gait function of stroke patients. *J Phys Ther Sci*. (2016) 28:3149–52. doi: 10.1589/jpts.28.3149
39. Ebrahimi A, Eftekhari E, Etemadifar M. Effects of whole body vibration on hormonal & functional indices in patients with multiple sclerosis. *Indian J Med Res*. (2015) 142:450–8. doi: 10.4103/0971-5916.169210
40. Furness TP, Maschette WE. Influence of whole body vibration platform frequency on neuromuscular performance of community-dwelling older adults. *J Strength Cond Res*. (2009) 23:1508–13. doi: 10.1519/JSC.0b013e3181a4e8f9
41. Gaßner H, Janzen A, Schwirtz A, Jansen P. Random whole body vibration over 5 weeks leads to effects similar to placebo: A controlled study in Parkinson's Disease. *Parkinson's Dis*. (2014) 2014:1–9. doi: 10.1155/2014/386495
42. Gerhardt F, Dumitrescu D, Gärtner C, Beccard R, Viethen T, Kramer T, et al. Oscillatory whole-body vibration improves exercise capacity and physical performance in pulmonary arterial hypertension: a randomised clinical study. *Heart*. (2017) 103:592–8. doi: 10.1136/heartjnl-2016-309852
43. Gloeckl R, Heinzelmann I, Seeberg S, Damisch T, Hitzl W, Kenn K. Effects of complementary whole-body vibration training in patients after lung transplantation: a randomized, controlled trial. *J Heart Lung Transpl*. (2015) 34:1455–61. doi: 10.1016/j.healun.2015.07.002
44. Gómez-Cabello A, González-Agüero A, Ara I, Casajús JA, Vicente-Rodríguez G. Effects of a short-term whole body vibration intervention on physical fitness in elderly people. *Maturitas*. (2013) 74:276–8. doi: 10.1016/j.maturitas.2012.12.008
45. Goudarzian M, Ghavi S, Shariat A, Shirvani H, Rahimi M. Effects of whole body vibration training and mental training on mobility, neuromuscular performance, and muscle strength in older men. *J Exer Rehabil*. (2017) 13:573–80. doi: 10.12965/jer.1735024.512
46. Höglér W, Scott J, Bishop N, Arundel P, Nightingale P, Mughal MZ, et al. The effect of whole body vibration training on bone and muscle function in children with osteogenesis imperfecta. *J Clin Endocrinol Metabol*. (2017) 102:2734–43. doi: 10.1210/enc.2017-00275
47. In T, Jung K, Lee M-G, Cho H. Whole-body vibration improves ankle spasticity, balance, and walking ability in individuals with incomplete cervical spinal cord injury. *NeuroRehabilitation*. (2018) 42:491–7. doi: 10.3233/NRE-172333
48. Iwamoto J, Sato Y, Takeda T, Matsumoto H. Whole body vibration exercise improves body balance and walking velocity in postmenopausal osteoporotic women treated with alendronate: Galileo and Alendronate Intervention Trial (GAIT). *J Musculoskelet Neuronal Interact*. (2012) 12:136–43.
49. Johnson AW, Myrer JW, Hunter I, Feland JB, Hopkins JT, Draper DO, et al. Whole-body vibration strengthening compared to traditional strengthening during physical therapy in individuals with total knee arthroplasty. *Physiother Theor Prac*. (2010) 26:215–25. doi: 10.3109/09593980902967196
50. Lam FM, Chan PF, Liao L, Woo J, Hui E, Lai CW, et al. Effects of whole-body vibration on balance and mobility in institutionalized older adults: a randomized controlled trial. *Clin Rehabil*. (2018) 32:462–72. doi: 10.1177/0269215517733525
51. Lau RWK, Yip SP, Pang MYC. Whole-body vibration has no effect on neuromotor function and falls in chronic stroke. *Med Sci Sports Exerc*. (2012) 44:1409–1418. doi: 10.1249/MSS.0b013e31824e4f8c
52. Lee B-K, Chon S-C. Effect of whole body vibration training on mobility in children with cerebral palsy: a randomized controlled experimenter-blinded study. *Clin Rehabil*. (2013) 27:599–607. doi: 10.1177/0269215512470673
53. Lee K, Lee S, Song C. Whole-Body vibration training improves balance, muscle strength and glycosylated hemoglobin in elderly patients with diabetic neuropathy. *Tohoku J Exp Med*. (2013) 231:305–14. doi: 10.1620/tjem.231.305
54. Liao L-R, Ng GYE, Jones AYM, Huang M-Z, Pang MYC. Whole-body vibration intensifies in chronic stroke: a randomized controlled trial. *Med Sci Sports Exerc*. (2016) 48:1227–38. doi: 10.1249/MSS.0000000000000909
55. Machado A, García-López D, González-Gallego J, Garatachea N. Whole-body vibration training increases muscle strength and mass in older women: a randomized-controlled trial: whole-body vibration in older women. *Scan J Med Sci Sports*. (2009) 20:200–7. doi: 10.1111/j.1600-0838.2009.00919.x
56. Mikhael M, Orr R, Amsen F, Greene D, Fiatarone Singh MA. Effect of standing posture during whole body vibration training on muscle morphology and function in older adults: a randomised controlled trial. *BMC Geriatr*. (2010) 10:74. doi: 10.1186/1471-2318-10-74
57. Pleguezuelos E, Pérez ME, Guirao L, Samitier B, Costea M, Ortega P, et al. Effects of whole body vibration training in patients with severe chronic obstructive pulmonary disease: whole body vibration in COPD. *Respirology*. (2013) 18:1028–34. doi: 10.1111/resp.12122
58. Raimundo AM, Gusi N, Tomas-Carus P. Fitness efficacy of vibratory exercise compared to walking in postmenopausal women. *Eur J Appl Physiol*. (2009) 106:741–8. doi: 10.1007/s00421-009-1067-9
59. Rees S, Murphy A, Watsford M. Effects of vibration exercise on muscle performance and mobility in an older population. *J Aging Phys Activ*. (2007) 15:367–81. doi: 10.1123/japa.15.4.367
60. Ruck J, Chabot G, Rauch F. Vibration treatment in cerebral palsy: A randomized controlled pilot study. *J Musculoskelet Neuronal Interact*. (2010) 10:77–83.
61. Salhi B, Malfait TJ, Van Maele G, Joos G, van Meerbeeck JP, Derom E. Effects of whole body vibration in patients with COPD. *COPD: J Chron Obst Pulm Dis*. (2015) 12:525–32. doi: 10.3109/15412555.2015.1008693
62. Santin-Medeiros F, Rey-López JP, Santos-Lozano A, Cristi-Montero CS, Garatachea Vallejo N. Effects of eight months of whole-body vibration



- training on the muscle mass and functional capacity of elderly women. *J Strength Cond Res.* (2015) 29:1863–1869. doi: 10.1519/JSC.0000000000000830
63. Simão AP, Avelar NC, Tossige-Gomes R, Neves CD, Mendonça VA, Miranda AS, et al. Functional performance and inflammatory cytokines after squat exercises and whole-body vibration in elderly individuals with knee osteoarthritis. *Arch Phys Med Rehabil.* (2012) 93:1692–700. doi: 10.1016/j.apmr.2012.04.017
  64. Sitjà-Rabert M, Martínez-Zapata MJ, Fort Vanmeerhaeghe A, Rey Abella F, Romero-Rodríguez D, Bonfill X. Effects of a whole body vibration (WBV) exercise intervention for institutionalized older people: a randomized, multicentre, parallel, clinical trial. *J Am Med Directors Assoc.* (2015) 16:125–31. doi: 10.1016/j.jamda.2014.07.018
  65. Spielmanns M, Boeselt T, Gloeckl R, Klutsch A, Fischer H, Polanski H, et al. Low-volume whole-body vibration training improves exercise capacity in subjects with mild to severe COPD. *Res. Care.* (2017) 62:315–23. doi: 10.4187/respcare.05154
  66. Sucuoglu H, Tuzun S, Akbaba YA, Uludag M, Gokpinar HH. Effect of whole-body vibration on balance using posturography and balance tests in postmenopausal women. *Am J Phy Med Rehabil.* (2015) 94:499–507. doi: 10.1097/PHM.0000000000000325
  67. van Nes Ilse JW, Latour H, Schils F, Meijer R, van Kuijk A, Geurts Alexander C, et al. Long-term effects of 6-week whole-body vibration on balance recovery and activities of daily living in the postacute phase of Stroke. *Stroke.* (2006) 37:2331–5. doi: 10.1161/01.STR.0000236494.62957.f3
  68. Wang P, Yang L, Liu C, Wei X, Yang X, Zhou Y, et al. Effects of whole body vibration exercise associated with quadriceps resistance exercise on functioning and quality of life in patients with knee osteoarthritis: a randomized controlled trial. *Clin Rehabil.* (2016) 30:1074–87. doi: 10.1177/0269215515607970
  69. Wei N, Pang MY, Ng SS, Ng GY. Optimal frequency/time combination of whole body vibration training for developing physical performance of people with sarcopenia: a randomized controlled trial. *Clin Rehabil.* (2017) 31:1313–21.
  70. Rauch F, Sievanen H, Boonen S, Cardinale M, Degens H, Felsenberg D, et al. Reporting whole-body vibration intervention studies: recommendations of the International Society of musculoskeletal and neuronal interactions. *J Musculoskelet Neuronal Interact.* (2010) 6:193–8.
  71. Spielmanns M, Gloeckl R, Gropp JM, Nell C, Koczulla AR, Boeselt T, et al. Whole-Body vibration training during a low frequency outpatient exercise training program in chronic obstructive pulmonary disease patients: a randomized, controlled trial. *J Clin Med Res.* (2017) 9:396–402. doi: 10.14740/jocmr2763w
  72. Choi W, Han D, Kim J, Lee S. Whole-body vibration combined with treadmill training improves walking performance in post-stroke patients: a randomized controlled trial. *Med Sci Monitor.* (2017) 23:4918–25. doi: 10.12659/MSM.904474
  73. Gloeckl R, Heinzelmann I, Bauerle S, Damm E, Schwedhelm A-L, Diril M, et al. Effects of whole body vibration in patients with chronic obstructive pulmonary disease – A randomized controlled trial. *Respiratory Med.* (2012) 106:75–83. doi: 10.1016/j.rmed.2011.10.021
  74. Park SY, Son WM, Kwon OS. Effects of whole body vibration training on body composition, skeletal muscle strength, and cardiovascular health. *J Exerc Rehabil.* (2015) 11:289–95. doi: 10.12965/jer.150254
  75. Papa EV, Dong X, Hassan M. Resistance training for activity limitations in older adults with skeletal muscle function deficits: a systematic review. *Clin Interv Aging.* (2017) 12:955–61. doi: 10.2147/CLIA.S104674
  76. Lee PG, Jackson EA, Richardson CR. Exercise prescriptions in older adults. *Am Fam Phys.* (2017) 95:425–432.
  77. Mora JC, Valencia WM. Exercise and older adults. *Clinics Geriatr Med.* (2018) 34:145–62. doi: 10.1016/j.cger.2017.08.007
  78. Tinetti, M. E. (2003). Clinical practice. Preventing falls in elderly persons. *N Engl J Med.* 348, 42–49. doi: 10.1056/NEJMcp020719
  79. Curcio F, Basile C, Liguori I, Della-Morte D, Gargiulo G, Galizia G, et al. Tinetti mobility test is related to muscle mass and strength in non-institutionalized elderly people. *Age.* (2016) 38:525–33. doi: 10.1007/s11357-016-9935-9
  80. Dunn A, Marsden DL, Nugent E, Van Vliet P, Spratt NJ, Attia J, et al. Protocol variations and six-minute walk test performance in stroke survivors: a systematic review with meta-analysis. *Stroke Res Treat.* (2015) 2015:484813. doi: 10.1155/2015/484813
  81. Bartholdy C, Juhl C, Christensen R, Lund H, Zhang W, Henriksen M. The role of muscle strengthening in exercise therapy for knee osteoarthritis: a systematic review and meta-regression analysis of randomized trials. *Semin Arthritis Rheum.* (2017) 47:9–21. doi: 10.1016/j.semarthrit.2017.03.007
  82. Gojanovic B and Henchoz Y. Whole-body vibration training: Metabolic cost of synchronous, side-alternating or no vibrations. *J Sports Sci.* (2012) 30:1397–403. doi: 10.1080/02640414.2012.710756
  83. Ritzmann R, Gollhofer A, Kramer A. The influence of vibration type, frequency, body position and additional load on the neuromuscular activity during whole body vibration. *Eur J Appl Physiol.* (2013) 113:1–11. doi: 10.1007/s00421-012-2402-0
  84. Chan K-S, Liu CW, Chen TW, Weng MC, Huang MH, Chen CH. Effects of a single session of whole body vibration on ankle plantarflexion spasticity and gait performance in patients with chronic stroke: a randomized controlled trial. *Clin Rehabil.* (2012) 26:1087–95. doi: 10.1177/0269215512446314
  85. Schuhfried O, Mittermaier C, Jovanovic T, Pieber K, Paternostro-Sluga T. Effects of whole-body vibration in patients with multiple sclerosis: a pilot study. *Clin Rehabil.* (2005) 19:834–42. doi: 10.1191/0269215505cr919oa
  86. Silva AT, Dias MPF, Calixto R, Carone AL, Martinez BB, Silva AM, et al. Acute effects of whole-body vibration on the motor function of patients with stroke: a randomized clinical trial. *Am J Phy Med Rehabil.* (2014) 93:310–9. doi: 10.1097/PHM.0000000000000042
  87. Noyes J, Booth A, Flemming K, Garside R, Harden A, Lewin S, et al. Cochrane qualitative and implementation methods group guidance paper 3: methods for assessing methodological limitations, data extraction and synthesis, and confidence in synthesized qualitative findings. *J Clin Epidemiol.* (2018) 97:49–58. doi: 10.1016/j.jclinepi.2017.06.020
  88. Higgins JP, Green, S. *Cochrane Handbook for Systematic Reviews of Interventions.* (2019) Available online at: <https://handbook-5-1.cochrane.org> (accessed March 26, 2019).
  89. Tinetti ME. Performance-oriented assessment of mobility problems in elderly patients. *J Am Geriatr Soc.* (1986) 34:119–26.
  90. ATS Statement. Guidelines for the Six-Minute Walk Test. *Am J Respir Crit Care Med.* (2002) 166:111–117. doi: 10.1164/ajrccm.166.1.at1102
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### **3.1.2. Effets immédiats des vibrations corps entier sur l'organisation posturale de l'initiation de la marche chez les jeunes adultes et les personnes âgées : étude randomisée contrôlée**

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# Acute Effects of Whole-Body Vibration on the Postural Organization of Gait Initiation in Young Adults and Elderly: A Randomized Sham Intervention Study

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Whole-body vibration (WBV) is a training method that exposes the entire body to mechanical oscillations while standing erect or seated on a vibrating platform. This method is nowadays commonly used by clinicians to improve specific motor outcomes in various sub-populations such as elderly and young healthy adults, either sedentary or well-trained. The present study investigated the effects of acute WBV application on the balance control mechanisms during gait initiation (GI) in young healthy adults and elderly. It was hypothesized that the balance control mechanisms at play during gait initiation may compensate each other in case one or several components are perturbed following acute WBV application, so that postural stability and/or motor performance can be maintained or even improved. It is further hypothesized that this capacity of adaptation is altered with aging. Main results showed that the effects of acute WBV application on the GI postural organization depended on the age of participants. Specifically, a positive effect was observed on dynamic stability in the young adults, while no effect was observed in the elderly. An increased stance leg stiffness was also observed in the young adults only. The positive effect of WBV on dynamic stability was ascribed to an increase in the mediolateral amplitude of “anticipatory postural adjustments” following WBV application, which did overcompensate the potentially destabilizing effect of the increased stance leg stiffness. In elderly, no such anticipatory (nor corrective) postural adaptation was required since acute WBV application did not elicit any change in the stance leg stiffness. These results suggest that WBV application may be effective in improving dynamic stability but at the condition that participants are able to develop adaptive changes in balance control mechanisms, as did the young adults. Globally, these findings are thus in agreement with the hypothesis that balance control mechanisms are interdependent within the postural system, i.e., they may compensate each other in case one component (here the leg stiffness) is perturbed.

**Keywords:** whole-body vibration (WBV), gait initiation, elderly, young adults, anticipatory postural adjustment (APA)

## INTRODUCTION

Whole-body vibration (WBV) is a training method that exposes the entire body to mechanical oscillations while standing erect or seated on a vibrating platform. This method is nowadays commonly used by clinicians to improve specific motor outcomes in various sub-populations such as elderly and young healthy adults, either sedentary or well-trained. Specifically, it has been shown that WBV application might be beneficial to numerous lower limbs motor outcomes, such as endurance, power, strength, neuromuscular activity, flexibility, and knee stability [for young healthy subjects, see for example (1–7); for elderly, see for example (8–13), for review see (14)].

Balance and locomotion might also be improved by WBV application, as quantified (i) with center of pressure (COP) measures in static upright condition [for young healthy adults, see for example (3, 15, 16); for elderly, see for example (17–20)], and (ii) with commonly-used clinical tests such as the single-leg-stance test, the Tinetti test, the Timed-Up-and-Go test, the Berg Balance Scale, and the Limits of stability. However, it is well-recognized that these classical clinical tests are subjective, show ceiling effects, and are usually not responsive enough to measure small progresses or deteriorations in a subject's ability to balance (21, 22). Mancini and Horak (22) further stressed that the greatest limitation of this clinical approach to rating balance is that it cannot specify what type of balance problem a subject suffers in order to direct a treatment. Upright balance maintenance during locomotor tasks is indeed a highly complex task since it necessitates the coordination between many complementary postural mechanisms, including dynamic postural phenomena before the swing foot clearance (23–25), the regulation of stance leg stiffness (26), the swing foot placement on the support surface (27), the braking of center of mass fall (28–30) etc. The investigation of these mechanisms [cf. description below; see also (31) for a recent review] requires an in-depth biomechanical analysis with force plate recordings. Such analysis is not provided in the classical clinical tests reported above. In fact, the classical clinical test, as Timed Up and Go test for example, could be limited to determined individual risk of falls in elderly (32). Furthermore, Topper et al. (33) show that falls happened most frequently in elderly subjects scoring poorly on clinical tests emphasizing transfer of quasi-static to dynamic situations.

Gait initiation is a functional task that is classically used in the literature as an experimental paradigm to investigate how the central nervous system (CNS) controls balance in dynamical conditions in both healthy adults [e.g., (31–35) for recent articles] and pathological patients [e.g., (36–40) for recent articles]. Gait initiation appears to be the major biomechanical model, complementary to the classical clinical test, in order to predict postural disorders in elderly fallers and non-fallers (41–44). Therefore, it seems highly appropriate for an investigation on the effects of WBV on dynamic balance. Gait initiation corresponds to the transient period between static erect posture and steady state walking (23–25). It can be divided into two successive phases: a postural phase that precedes the onset of the swing heel off (the so called “anticipatory postural adjustments”,

APA) followed by an execution phase that ends at the time of swing foot contact (FC) with the ground.

During APA, the COP is shifted backward, creating the initial forward propulsive forces that are necessary to reach the intended center of mass (COM) velocity and step length (23, 25, 45). At the same time, the COP is shifted laterally toward the swing leg, which acts to propel the COM toward the stance leg (23, 26, 35, 46, 47). This anticipatory mediolateral postural dynamic acts to attenuate the COM fall toward the swing leg during the execution phase under gravity (26, 48, 49). It has therefore a stabilizing function. In case mediolateral APA are not sufficient to ensure stabilization, a strategy of lateral swing foot placement that enlarges the base of support size has been documented (27, 46, 47, 50, 51). The development of APA and the lateral swing foot placement during locomotor tasks are known to be dependent on somatosensory inputs from the lower limbs (52–55). The perturbation of these inputs by the applications of WBV might thus alter these two mechanisms and increase instability and/or reduce motor performance.

In addition to mediolateral APA and lateral swing foot placement, recent modeling study showed that increasing the stance leg stiffness along the mediolateral direction would result in a larger COM fall toward the swing leg side, thus leading to an increased mediolateral instability. The application of WBV may trigger a “tonic vibration reflex” (TVR) via the activation of muscle spindles of the lower limbs (56, 57). Although controversial (8, 58–60), this TVR might potentially be responsible for an increase in lower limbs stiffness and may thus have a negative impact on stability.

Globally, this brief literature review suggests that WBV may potentially perturb the balance control mechanisms involved in gait initiation and may therefore induce instability. This statement thus markedly contrasts with the results of the above-reported clinical tests, which rather suggest that WBV may be effective in improving relatively basic balance ability and locomotion. Thus, the purpose of this study is to clarify the effects of acute WBV application on the balance control mechanisms during gait initiation in young healthy adults and elderly. It is hypothesized that the balance control mechanisms at play during gait initiation may compensate each other in case one or several components are perturbed following acute WBV application, so that postural stability and/or motor performance can be maintained (or even improved). It is further hypothesized that this capacity of adaptation is altered with aging, with a consequent increased instability following WBV application.

## METHODS

### Participants

The study is a randomized investigation that includes 81 healthy adults (i.e., 41 young and 40 elderly). The non-probability convenience method is used, i.e., participants were randomly assigned to one of the four following treatment groups using the “envelope method” (see below): (1) 20 young adults (11 men and 9 women, mean age  $25.3 \pm 3.5$  years, height  $1.71 \pm 0.05$  m and body-mass  $71.4 \pm 5.3$  kg) were assigned to the Young WBV training group (YWBV); (2) 21 matched young adults

were assigned to the Young Sham Group (YSG) (11 men and 10 women, mean age  $26.6 \pm 4.2$  years, height  $1.74 \pm 0.04$  m and body-mass  $72.1 \pm 4.9$  kg); (3) 20 elderly (9 men and 11 women, mean age  $83.5 \pm 2.8$  years, height  $1.69 \pm 0.06$  m and body-mass  $70.4 \pm 4.3$  kg) were assigned to the Elderly WBV training group (EWBV) and (4) 20 matched elderly (9 men and 11 women, mean age  $84.2 \pm 3.7$  years, height  $1.71 \pm 0.07$  m and body-mass  $71.7 \pm 3.9$  kg) were assigned to the Elderly Sham Group (ESG; see **Table 1**). In the “envelope method,” each participant chooses a closed envelope in which his/her treatment group is indicated in a code form. The participant does not have access to this code but the experimenters do (single blinded condition).

The two WBV groups (young adults and elderly) received a single WBV training and the two sham groups received a placebo treatment (cf. below for description). As stated above, participants were blinded to their allocation group. Participants had no recent history of trauma, known metabolic disorders, inflammatory infectious arthropathies, bone malignancies or neurological disease. In addition, participants were all naïve about WBV treatment. They all gave written consent after having been informed of the nature and purpose of the experiment which was approved by local ethics committees from the CIAMS Research Unit, Equipe d'Accueil (EA) 4532. The study complied with the standards established by the Declaration of Helsinki and was assigned the following trial registration number: 2017-002539-40.

### Experimental Task and Conditions

Experiments took place in the Biomechanics laboratory of the Paris-Saclay University located within the Kremlin-Bicêtre Hospital (Paris, France). Physical conditions (room temperature and time of the day) were common to all treatment groups, and the same before/after the WBV treatment.

Participants initially stood barefoot in a natural upright posture on a force plate ( $0.9 \times 1.80$  m, AMTI, Watertown, USA) embedded at the beginning of a 6 m walkway track. The feet

were shoulder-width apart, with the arms alongside the trunk and the gaze directed forward to a small target at eye level (2 cm diameter, 5 m distant). The locations of the heel and big toe of each foot in the initial posture were marked with sections of adhesive tape placed on the force plate and were used as a visual reference on which participants positioned themselves under the supervision of the experimenters. From this initial posture, participants performed two series of 10 gait initiation trials: one series just before, and a second series immediately after a specific treatment (pre- and post-treatment conditions, respectively) depending on their group (WBV or sham). In both pre- and post-treatment conditions, participants initiated gait at a spontaneous velocity and at their own initiative following an auditory signal delivered by the experimenter, and then continued walking straight until the end of the track. Participants initiated gait with their preferred leg in all trials. One blank trial was provided in the pre-manipulation condition (not recorded) to ensure that the instructions were well-understood by the participant and that the material was operational. The rest time was  $\sim 10$  s between trials.

### WBV and Placebo Treatment

For both the WBV group and the sham group, participants stood upright on a vibration platform (referee Power Fitness Double D<sup>®</sup>) with knees flexed  $\sim 30^\circ$  and barefoot. During the treatment, knee angle was monitored on-line by both the subject him/herself and the experimenters thanks to a visual feedback provided by the signal of a monoaxial electrogoniometer (Biometrics<sup>®</sup>) fixed on the right leg. Participants were instructed to maintain the signal within a 25–35 range degrees.

For both young adults and elderly, the WBV application included four 45 s bouts of vibration, intercalated by 1 min of seated rest without vibration between bouts (61–63). Vibrations were delivered by the vibration platform at a frequency of 50 Hz [this frequency is known to enhance locomotor task and intralimb coordination (61)] with a vertical displacement of 2 mm and a 2 g acceleration.

The sham group received a treatment that followed exactly the same procedure as the WBV group except that no “real” vibrations were applied to participants. In order to maintain the most realistic placebo conditions as possible, participants were instructed by an experimenter that they could not physically feel the vibrations because they were too small to be sensed.

### Mechanical Model

The same mechanical model as in a previous study was used to determine the stance leg stiffness along the mediolateral direction during the execution phase of gait initiation [for details on this model and the related equations of motion, see (26)]. In brief, the human body was modeled as a single conic inverted pendulum which rotates around a fixed point corresponding to the stance ankle. It was considered that the COM falls laterally under the influence of two forces: the gravity force  $P = mg$  (where  $m$  is the mass of the whole body, and  $g$  is the gravitational acceleration) and an elastic restoring mediolateral force  $T$  (in Newtons) that reflects active muscular control of the movement (64, 65), with  $T = k|yM|$  where  $k$  is the stance leg stiffness (Newtons/meter)

**TABLE 1** | Anthropometric data of subjects in both WBV and sham groups (young adults and elderly).

Group	Variables	WBV group	Sham group	P-values
Young adults (n = 41)	Women (n)	9	10	NS
	Men (n)	11	11	NS
	Age (years)	$25.30 \pm 3.50$	$26.6 \pm 4.2$	NS
	Height (m)	$1.71 \pm 0.05$	$1.74 \pm 0.04$	NS
Elderly (n = 40)	Body mass (kg)	$71.40 \pm 5.30$	$72.10 \pm 4.90$	NS
	Women (n)	11	11	NS
	Men (n)	9	9	NS
	Age (years)	$83.50 \pm 2.80$	$84.20 \pm 3.70$	NS
	Height (m)	$1.69 \pm 0.06$	$1.71 \pm 0.07$	NS
	Body mass (kg)	$70.40 \pm 4.30$	$71.70 \pm 3.90$	NS

Reported values are means  $\pm$  standard deviation.

NS, non-significant difference.

and  $|y_M|$  is the absolute value of the mediolateral COM shift (in meters), systematically oriented toward the swing leg (positive values) during the execution phase. The initial position and velocity of the cone corresponded to the position and velocity of the participant's COM at toe off.

### Raw Data

Force-plate data were low-pass filtered using a second order Butterworth filter with a 10 Hz (30) cut-off frequency. The mediolateral ( $y_p$ ) and anteroposterior ( $x_p$ ) coordinates of the COP (in meters) were computed from force-plate data as follows:

$$y_p = \frac{M_x + F_y \times dz}{F_z} \quad (1)$$

$$x_p = \frac{-M_y + F_x \times dz}{F_z} \quad (2)$$

where  $M_x$  and  $M_y$  are the moments (in Newtons.meters) around the anteroposterior and mediolateral axes (at the force-plate origin), respectively;  $F_x$ ,  $F_y$ , and  $F_z$  (in Newtons) are the mediolateral, anteroposterior, and vertical ground reaction forces, respectively; and  $dz$  (in meters) is the distance between the surface of the force-plate and its origin which is located below this surface.

Instantaneous COM acceleration along the anteroposterior and mediolateral axes was determined from the ground reaction forces according to the Newton's second law. COM velocity and displacement were computed by successive numerical integrations of COM acceleration using the rectangles method (66) and integration constants equal to zero, i.e., initial velocity and displacement null (23).

### Experimental Dependent Variables

The biomechanical traces and the representations of the main experimental variables obtained for one representative young subject of the WBV group initiating gait (one trial) in the pre-treatment condition and post-treatment condition are presented in the **Figure 2**. The following instants were determined from the biomechanical traces: gait initiation onset ( $t_0$ ), swing heel-off, swing toe-off, swing foot-contact, and rear foot-off. These instants were determined from force-plate data (23, 67, 68). Two  $t_0$  times were estimated, one for the mediolateral axis and one for the anteroposterior axis. The  $t_0$  times corresponded to the instants when the mediolateral or anteroposterior COP trace deviated 2.5 standard deviations from its baseline value.

The mediolateral and anteroposterior COM position in the initial upright static posture were estimated by averaging the COP position during the 250 ms period preceding the "all set" signal. Gait initiation was divided into APA (from  $t_0$  to heel-off), swing foot-lift (from heel-off to toe-off), and execution phase (from toe-off to foot-contact). The duration of APA along the mediolateral and anteroposterior axes were computed separately, because the  $t_0$  times for these two axes do not necessarily occur simultaneously. The amplitude of APA was characterized by the peaks of the backward and mediolateral COP shift obtained during the APA time window. COM velocity and displacement along the mediolateral and anteroposterior axes were quantified at heel-off and foot-contact.

The spatiotemporal features of gait performance during the execution phase included: the swing phase duration, the anteroposterior COM velocity at foot-contact (progression velocity), and the step length. Step length corresponds to the difference between the most backward COP position during gait initiation and the COP position at the time of rear foot-off (69). Rear foot off time was marked with the mediolateral COP trace.

The spatiotemporal features of postural stability during the execution phase included: the step width, the ML COM position at FC, the ML COM velocity at FC and the margin of stability (MOS). An adaptation of the MOS introduced by Hof et al. (70) was used to quantify the mediolateral dynamic stability at foot-contact (thereafter referred to as "stability"). The MOS corresponds to the difference between the mediolateral boundary of the base of support (BOS<sub>ymax</sub>, in meters) and the mediolateral position of the "extrapolated COM" at swing foot-contact ( $y_{COMFC}$ , in meters). Thus:

$$MOS = BOS_{ymax} - y_{COMFC} \quad (3)$$

Participants systematically landed on the force-plate with the swing heel first, then the toe. Under such foot landing strategy, it is known that BOS<sub>ymax</sub> could be estimated with the mediolateral COP position at the time of rear foot-off (69, 71). Based on the study by Hof et al. (70) and the results from our previous studies (26, 27, 69, 71, 72) the mediolateral position of the extrapolated COM at foot-contact ( $y_{COMFC}$ ) was calculated as follows:

$$y_{COMFC} = y_{MFC} + \frac{y'_{MFC}}{\omega_0} \quad (4)$$

where  $y_{MFC}$  and  $y'_{MFC}$  are respectively the mediolateral COM position (in meters) and velocity at foot-contact (in meters/second), and  $\omega_0$  is the eigenfrequency of the body (Hz), modeled as an inverted pendulum and calculated as follows:

$$\omega_0 = \sqrt{\frac{g}{l}} \quad (5)$$

where  $g = 9.81 \text{ m/s}^2$  is the gravitational acceleration and  $l$  is the length of the inverted pendulum, which in this study corresponded to 57.5% of body height (73). Mediolateral dynamic stability at foot-contact is preserved on the condition that  $y_{COMFC}$  is within BOS<sub>ymax</sub>, which corresponds to a positive MOS. A negative MOS indicates a mediolateral instability and implies that a corrective action is required to maintain balance. Hence, variables related to the mediolateral stability included the MOS and its components, i.e., step width, mediolateral COM velocity and position at swing foot-contact. Step width corresponds to the difference between the most lateral position of the mediolateral COP trace obtained during the first plateau of the trace and the mediolateral COP position at the time of rear foot-off (69).

### Statistics

Mean values and standard deviations were calculated for each variable in each condition. The normality of data was checked

using the Kolmogorov–Smirnov test and the homogeneity of variances was checked using the Bartlett test. Repeated measures (RM) ANOVAs with the treatment (pre-treatment condition vs. post-treatment condition) as within subject factor and the group (sham group vs. WBV group) as between subjects' factor were used on each experimental variable. Because this research focuses on the effect of WBV application on gait initiation in young and elderly adults, merely than on the effects of age on gait initiation, the RM ANOVAs were conducted for young and elderly adults taken separately. When necessary, a significant outcome was followed-up with the Tukey *post-hoc* test. Finally, linear correlation between variables was assessed with the Pearson coefficient  $r$ . The threshold of significance was set at  $p < 0.05$ .

## RESULTS

### Biomechanical Traces

The time course of the biomechanical traces was globally similar in the two groups (young adults and elderly) and in the different treatment conditions (WBV and sham). The traces obtained in a representative young adult subject, before and after the WBV treatment, are reported in the **Figure 1**.

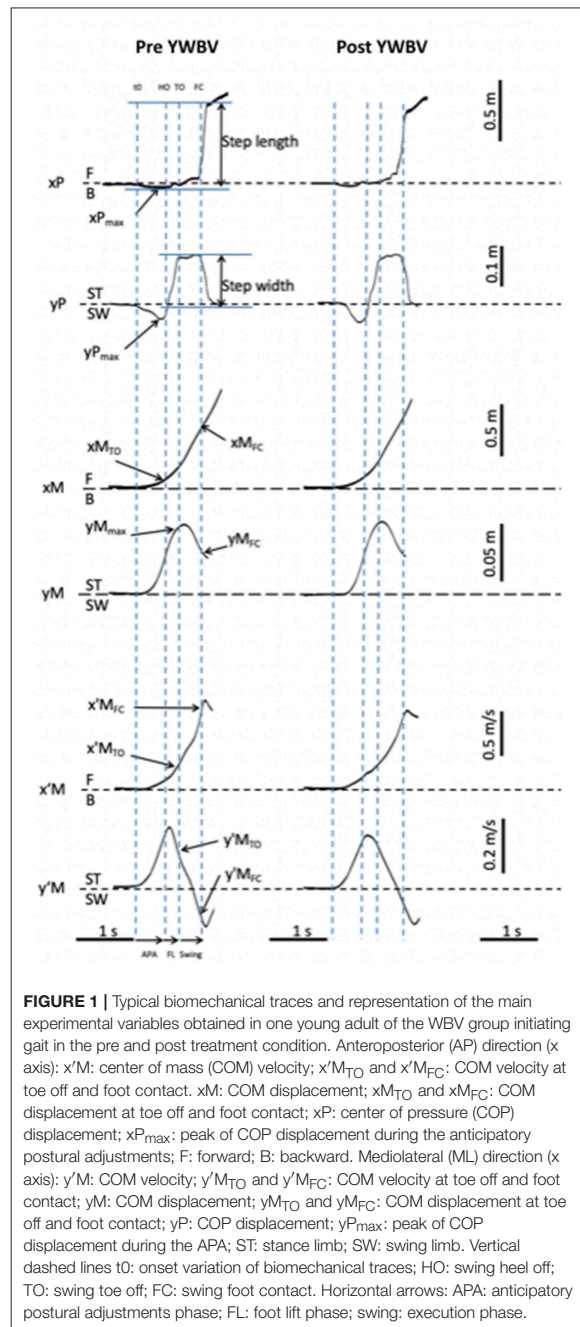
As can be seen from this figure, swing heel off was systematically preceded by postural dynamics that corresponded to APA. During these APA, the COP displacement reached a peak value in a backward direction (see the negative variation of the  $xP$  trace) and toward the swing leg side (negative variation of the  $yP$  trace), while the COM velocity was directed forwards (positive variation of the  $x'M$  trace) and toward the stance leg side (positive variation of the  $y'M$  trace). The mediolateral COM velocity trace reached a first peak value toward the stance leg side at around heel off. This trace, then fell toward the swing leg side and a second peak value toward this side was reached a few milliseconds after foot contact. The anteroposterior COM velocity increased progressively until it reached a peak value a few milliseconds after swing foot contact. The mediolateral COM shift trace was bell-shaped and reached a peak value toward the stance leg side at the beginning of the swing phase, while the anteroposterior COM was continuously shifted forward (see **Figure 2**). Differences across the conditions are reported in the paragraphs below.

### Initial Posture, Anticipatory Postural Adjustments, and Foot-Lift Phase

In both the young adults and the elderly, no baseline differences were found between the WBV and sham groups ( $p > 0.05$ ) for all the variables (see **Table 2**).

In both the young adults and the elderly, statistical analysis showed that there was no significant effect of the group (sham group vs. WBV group) or treatment (pre-treatment condition vs. post-treatment condition) nor any significant interaction between these two factors on the initial COM position along the mediolateral or anteroposterior direction.

In the elderly, there was also no significant effect of the group or treatment nor any significant interaction between these two factors on the spatiotemporal parameters of APA along



**FIGURE 1** | Typical biomechanical traces and representation of the main experimental variables obtained in one young adult of the WBV group initiating gait in the pre and post treatment condition. Anteroposterior (AP) direction (x axis):  $x'M$ : center of mass (COM) velocity;  $x'M_{TO}$  and  $x'M_{FC}$ : COM velocity at toe off and foot contact;  $xM$ : COM displacement;  $xM_{TO}$  and  $xM_{FC}$ : COM displacement at toe off and foot contact;  $xP$ : center of pressure (COP) displacement;  $xP_{max}$ : peak of COP displacement during the anticipatory postural adjustments; F: forward; B: backward. Mediolateral (ML) direction (x axis):  $y'M$ : COM velocity;  $y'M_{TO}$  and  $y'M_{FC}$ : COM velocity at toe off and foot contact;  $yM$ : COM displacement;  $yM_{TO}$  and  $yM_{FC}$ : COM displacement at toe off and foot contact;  $yP$ : COP displacement;  $yP_{max}$ : peak of COP displacement during the APA; ST: stance limb; SW: swing limb. Vertical dashed lines t0: onset variation of biomechanical traces; HO: swing heel off; TO: swing toe off; FC: swing foot contact. Horizontal arrows: APA: anticipatory postural adjustments phase; FL: foot lift phase; swing: execution phase.

the mediolateral or anteroposterior direction (APA duration, peak COP shift, COM shift, and velocity at heel-off), nor on the swing foot-lift phase duration. In contrast, in the young adults, there was a significant effect of the treatment [ $F_{(1,39)} = 17.94$ ,  $p < 0.001$ ] on the mediolateral peak of anticipatory

COP shift, along with a significant group X treatment interaction [ $F_{(1,39)} = 8.83, p < 0.01$ ]. Similarly, there was a significant effect of the treatment [ $F_{(1,39)} = 8.89, p < 0.01$ ], along with a significant group X treatment interaction [ $F_{(1,39)} = 4.38, p = 0.04$ ] on the mediolateral COM velocity at heel-off. *Post-hoc* analysis further revealed that these two interactions could be ascribed to the result that none of these two APA related-parameters were significantly different between the pre and post-treatment condition for the sham group, while both were significantly larger in the post-treatment condition than in the pre-treatment condition for the WBV group ( $p < 0.001$  for the mediolateral peak of anticipatory COP shift

and  $p < 0.05$  for the mediolateral COM velocity at heel-off; cf. **Table 2**).

### Motor Performance

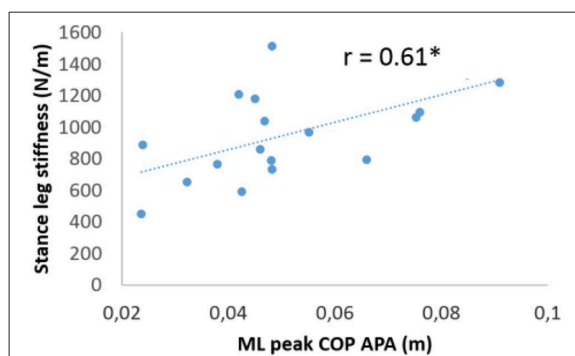
In both the young adults and the elderly, no baseline differences were found between the WBV and sham groups ( $p > 0.05$ ) for all the variables (see **Table 3**).

In both the young adults and the elderly, statistical analysis showed that there was no significant main effect of the group or treatment nor significant interaction between these two factors on the execution phase duration, step length, and progression velocity at foot contact.

### Stability

In both the young adults and the elderly, no baseline differences were found between the WBV and sham groups ( $p > 0.05$ ) for all the variables (cf. **Table 3**).

In the elderly, statistical analysis showed that there was no significant effect of the group or treatment nor any significant interaction between these two factors on the MOS and related components (i.e., step width, mediolateral COM position, and velocity at foot-contact). In contrast, in the young adults, there was a significant effect of the treatment, along with a significant group X treatment interaction on the following variables: MOS [ $F_{(1,39)} = 5.14, p = 0.03$ ;  $F_{(1,39)} = 5.62, p = 0.02$ , respectively], mediolateral COM position [ $F_{(1,39)} = 7.32, p = 0.01, F_{(1,39)} = 5.19, p = 0.03$ , respectively], and mediolateral COM velocity at foot-contact [ $F_{(1,39)} = 10.14, p < 0.01, F_{(1,39)} = 4.51, p = 0.04$ ; respectively]. *Post-hoc* analysis further revealed that these interactions could be ascribed to the result that none of these variables were significantly different between the pre and the post-treatment condition for



**FIGURE 2 |** Linear regression between normalized medio-lateral peak anticipatory center-of-pressure (COP) shift and the normalized medio-lateral stance leg stiffness in the WBV condition. Each point represents the average value of subjects in the WBV condition ( $r = 0.61, p < 0.01$ ).

**TABLE 2 |** Comparison of APA and foot lift parameters between the pre- and post-treatment condition in the WBV and sham group (young adults and elderly).

	APA and Foot lift phase Variables	WBV group		Sham group		Group X treatment interaction P-values
		Pre-treatment	Post-treatment	Pre-treatment	Post-treatment	
Young adults	AP COP shift (cm)	4.94 ± 1.51	5.29 ± 1.23	5.44 ± 1.37	5.52 ± 1.41	NS
	ML COP shift(cm)	4.61 ± 2.17	5.35 ± 2.09***	4.99 ± 1.27	5.12 ± 1.43	<0.001
	AP APA duration (s)	0.05 ± 0.11	0.47 ± 0.01	0.51 ± 0.10	0.50 ± 0.09	NS
	ML APA duration (s)	0.45 ± 0.08	0.47 ± 0.09	0.49 ± 0.01	0.45 ± 0.07	NS
	Foot lift duration (s)	0.12 ± 0.02	0.12 ± 0.03	0.14 ± 0.04	0.13 ± 0.04	NS
	AP COM velocity at HO (m/s)	0.21 ± 0.04	0.22 ± 0.05	0.23 ± 0.04	0.22 ± 0.04	NS
Elderly	ML COM velocity at HO (m/s)	0.14 ± 0.04	0.15 ± 0.04**	0.15 ± 0.02	0.15 ± 0.03	0.04
	AP COP shift (cm)	4.53 ± 1.23	5.03 ± 1.32	4.69 ± 0.64	4.95 ± 1.27	NS
	ML COP shift(cm)	4.21 ± 1.55	4.48 ± 1.63	4.85 ± 1.53	5.05 ± 1.88	NS
	AP APA duration (s)	0.40 ± 0.09	0.36 ± 0.06	0.38 ± 0.06	0.36 ± 0.05	NS
	ML APA duration (s)	0.39 ± 0.09	0.37 ± 0.05	0.36 ± 0.05	0.37 ± 0.05	NS
	Foot lift duration (s)	0.14 ± 0.02	0.14 ± 0.02	0.14 ± 0.02	0.14 ± 0.04	NS
	AP COM velocity at HO (m/s)	0.15 ± 0.05	0.16 ± 0.05	0.14 ± 0.03	0.15 ± 0.05	NS
	ML COM velocity at HO (m/s)	0.11 ± 0.04	0.11 ± 0.03	0.12 ± 0.04	0.12 ± 0.04	NS

Reported values are means ± standard deviation. APA, anticipatory postural adjustments; COM, center of mass; COP, center of pressure; AP, anteroposterior; ML, mediolateral; HO, heel off; NS, non-significant interaction. \*, \*\*, \*\*\*Significant difference between the pre and post-treatment condition as revealed with the Tukey post-hoc test, with  $P < 0.05$ ,  $P < 0.01$ , and  $P < 0.001$ , respectively.



**TABLE 3** | Comparison of motor performance, stability and stiffness parameters between the pre- and post-treatment condition in the WBV and sham group (young adults and elderly).

Motor performance, stability and stiffness		WBV group		Sham group		Group X treatment interaction
		Before	After	Before	After	
Young adults	Swing phase duration (s)	0.12 ± 0.02	0.13 ± 0.03	0.14 ± 0.04	0.14 ± 0.04	NS
	Step length (cm)	67.94 ± 6.43	67.89 ± 6.74	70.32 ± 8.61	70.68 ± 7.45	NS
	AP COM velocity at FC (m/s)	0.96 ± 0.01	0.98 ± 0.09	1.01 ± 0.12	1.02 ± 0.11	NS
	Step width (cm)	17.11 ± 4.03	17.07 ± 4.09	18.12 ± 4.57	17.52 ± 4.19	NS
	ML COM position at FC (cm)	3.56 ± 2.80	4.52 ± 2.11**	3.85 ± 2.27	3.93 ± 2.31	0.03
	ML COM velocity at FC (m/s)	-0.14 ± 0.05	-0.12 ± 0.04**	-0.15 ± 0.03	-0.14 ± 0.03	0.04
	Margin of stability (cm)	4.74 ± 3.43	5.76 ± 2.56*	5.79 ± 2.65	5.77 ± 2.48	0.02
	Stiffness	835.10 ± 355.40	975.62 ± 290.11*	973.32 ± 293.83	972.98 ± 328.87	0.04
Elderly	Swing phase duration (s)	0.14 ± 0.02	0.14 ± 0.04	0.14 ± 0.02	0.14 ± 0.04	NS
	Step length (cm)	52.74 ± 10.19	53.89 ± 8.35	52.74 ± 10.32	54.39 ± 9.20	NS
	AP COM velocity at FC (m/s)	0.74 ± 0.17	0.78 ± 0.11	0.76 ± 0.14	0.79 ± 0.11	NS
	Step width (cm)	16.84 ± 2.44	17.17 ± 2.77	17.49 ± 4.42	18.10 ± 4.96	NS
	ML COM position at FC (cm)	6.32 ± 5.19	5.98 ± 4.84	6.38 ± 6.34	6.11 ± 5.66	NS
	ML COM velocity at FC (m/s)	-0.13 ± 0.03	-0.14 ± 0.03	-0.14 ± 0.04	-0.15 ± 0.05	NS
	Margin of stability (cm)	5.98 ± 1.72	5.18 ± 2.21	6.38 ± 2.24	5.59 ± 2.58	NS
	Stiffness	1061.90 ± 145.99	1006.45 ± 221.21	1208.88 ± 389.14	1208.33 ± 508.51	NS

Reported values are means ± standard deviation. COM, center of mass; COP, center of pressure; AP, anteroposterior; ML, mediolateral; FC, foot contact; NS, non-significant interaction. \*, \*\*, \*\*\*Significant difference between the pre and post-treatment condition as revealed with the Tukey post-hoc test, with  $P < 0.05$ ,  $P < 0.01$ , and  $P < 0.001$ , respectively.

the sham group, while they were statistically different for the WBV group. Specifically, the mediolateral COM position ( $p < 0.01$ ) and the mediolateral COM velocity at foot-contact ( $p < 0.01$ ) were both significantly smaller in the post-treatment condition than in the pre-treatment condition for the WBV group, while the MOS was significantly larger ( $p < 0.05$ ; cf. Table 3).

### Stance Leg Stiffness

In both the young adults and the elderly, no baseline difference was found between the WBV and sham groups ( $p > 0.05$ ; cf. Table 3). In the elderly, statistical analysis showed that there was no significant effect of the group or treatment nor any significant interaction between these two factors on the leg stiffness. In contrast, in the young adults, there was a significant effect of the treatment [ $F_{(1,39)} = 4.15$ ,  $p < 0.05$ ] along with a significant group X treatment interaction [ $F_{(1,39)} = 4.19$ ,  $p = 0.04$ ] on this variable. *Post-hoc* analysis further revealed that this interaction could be ascribed to the result that leg stiffness was not significantly different between the pre and the post-treatment condition for the sham group, while it was statistically larger for the WBV group ( $p < 0.05$ ). In addition, results showed that, in the WBV group post-treatment condition, the stance leg stiffness in young adults was significantly correlated with the amplitude of mediolateral APA ( $r = 0.61$ ,  $p < 0.01$ ), i.e., the larger the leg stiffness, the larger the COP shift (Figure 2).

## DISCUSSION

In the paragraph that follows, the coherence of biomechanical data and their implication in terms of advanced postural control knowledge in young adults and elderly are discussed.

### Effect of Acute WBV Application on APA in Young Adults

In young adults, results showed that the amplitude of the mediolateral APA, expressed in terms of peak COP shift, was increased in the WBV group post-treatment. The difference between the pre- and the post-treatment condition reached up to 16.1%, which may be considered as clinically significant. The result that no difference was observed between the pre-treatment and the post-treatment condition in the sham group discards a placebo or a practice effect due to the repetition of the task. The increased mediolateral APA could neither be ascribed to an increase in the progression velocity (27) or to an increase in the duration of the execution phase of gait initiation (26, 46, 47, 72). Results indeed showed that there was no significant effect of the WBV application on these two parameters. In contrast to the mediolateral APA amplitude, the mediolateral APA duration did not change in the WBV group post-treatment. Participants were therefore able to generate larger APA within the same duration, i.e., the APA efficacy was increased post-treatment. In line with this statement, results showed that the mediolateral COM velocity at the time of swing foot-off, which is known to depend on

the spatiotemporal parameters of APA (27, 72), was larger following the WBV application. As this initial COM velocity is directed toward the stance leg, it acts to minimize the COM fall in the opposite direction (i.e., toward the swing leg) during the execution phase under the gravity effect (26, 49). The stabilizing function of APA was thus improved by acute WBV application.

The spatiotemporal parameters of APA are also known to be dependent on the somatosensory inputs arising from the lower limbs and trunk (52–55, 74–78). One can therefore wonder whether the positive effect on APA amplitude reported in this study could be ascribed to a change in the somatosensory inputs induced by the WBV application. While the current literature on healthy subjects repeatedly reported a detrimental effect of acute WBV application on the cutaneous lower limbs sensitivity (16, 79–81), it seems that acute WBV application does not influence lower limbs and trunk proprioception (17, 80–83); note that a detrimental effect on trunk proprioception was reported by Li et al. (84). Now, plantar sensitivity and limb proprioception were not investigated in the present study. However, an alteration of these inputs in the WBV group post-treatment (if any) may likely not be responsible for the changes in the mediolateral APA amplitude reported in the present study. For example, Lin and Yang (85) showed that the amplitude and the duration of mediolateral APA for gait initiation were both reduced (and not increased or unchanged, respectively, as in the present study) following plantar desensitization induced by cold water immersion. A decrease in the mediolateral APA for stepping was also reported when the ankle muscles acting in the mediolateral direction were vibrated to induce alteration of the proprioceptive afferent inflow from Ia fibers (53). In the present study, participants were exposed to WBV but, because of the standing posture with the feet on the vibrating force-plate, the ankle muscles were strongly submitted to the vibrations as in this latter study. The authors proposed that the proprioceptive information induced by vibration and afferent inflow related to body movement exaggerated sense of movement at the ankle. This biased perception led to a decrease in the APA spatiotemporal features. Similarly, a decrease in the postural response during balance recovery following cutaneous and muscular deafferentation experimentally induced by foot anesthesia and leg ischemia, respectively, was reported by Thoumie and Do (77). Finally, proprioceptive perturbation induced by experimental pain applied to the *tibialis anterior* (86) or acute fatigue of the same ankle muscles (78) were both shown to induce a decrease in the APA amplitude for gait initiation.

Globally, the above mentioned studies suggest that the increased mediolateral APA observed in the present research could probably not be ascribed to an (eventual) alteration of the somatosensory inputs induced by acute WBV application. Although not measured in these latter studies, it is possible that a major difference between these studies and the present one lays in the changes in the leg stiffness post-treatment. The stance leg stiffness was indeed increased by 16.8% during the execution phase of gait initiation following the WBV application, which may not be the case in the above mentioned

studies due to the different treatments. In the following paragraph, it is argued that this change in the mechanical feature of the stance leg might be responsible for the increased mediolateral APA.

### Effect of Acute WBV Application on Stance Leg Stiffness and Relationship With APA and Stability in Young Adults

As for APA, the increased stance leg stiffness observed in the WBV group post-treatment could not be ascribed to a placebo or a practice effect since no change in this parameter could be detected in the sham group pre-treatment vs. post-treatment. This increased leg stiffness might possibly be related to a “TVR” induced by the activation of leg muscle spindles sensitive to vibrations (56, 57). The effect of WBV on limb stiffness is however controversial in the literature. In human, no effect of acute WBV on stiffness has been reported on patellar tendon (87), hamstring, quadriceps (88), and triceps surae (89) and in lower limbs during hopping (90). In contrast, WBV has been shown to increase area and stiffness of the flexor carpi ulnaris tendon in rats (91) and in the Achille tendon of older women (8). Colson and Petit (90) concluded that further studies should be undertaken to ascertain the effectiveness of WBV on lower limbs stiffness. It is possible that the effect of acute WBV application on muscle stiffness depends on the task to be done and the muscle tested (and particularly its content in spindles), but also on the mechanical features of the vibrations (amplitude, frequency, duration).

Whatever its origin, previous modeling study showed that increasing stance leg stiffness has the potential to induce instability during gait initiation (26). The current results showed that acute WBV application induced two opposite effects on postural stabilization: a *stabilizing* effect via an increase in the mediolateral APA amplitude, and a *destabilizing* effect via an increase in stance leg stiffness. Results showed that the combined action of these two opposite effects resulted in an increased stability during gait initiation, as assessed with the larger mediolateral margin of stability post-treatment. This increase reached 21.4%, which can be considered as clinically relevant. This finding suggests that the negative effect of WBV application on leg stiffness and on related-stability was (over)compensated by the larger mediolateral APA. Furthermore, results showed that these two balance control parameters were positively correlated, i.e., the larger the stance leg stiffness, the larger the mediolateral APA amplitude. Based on these findings, the authors propose that the increased stance leg stiffness induced by the WBV application was taken into account in the programming of the mediolateral APA. In this hypothesis, WBV application would induce a postural constraint in the form of an increased stance leg stiffness requiring motor adaptation to maintain stability. This interpretation is in agreement with the hypothesis that balance control mechanisms are interdependent within the postural system, i.e., they may compensate each other in case one component (here the leg stiffness) is perturbed (31, 92–94). This interdependence would be necessary to maintain an

optimal control of stability in situations where a constraint is applied to the postural system. This statement is supported by recent studies. For example, during stepping over an obstacle in reaction to sudden force-plate translation, Zettel et al. (46, 47) reported that the drastic reduction of the mediolateral APA amplitude due to the urgency to clear the swing foot from the support surface did not result in a greater instability because it was compensated by larger step width, which corresponds to another form of balance control mechanism. In the same vein, Artico et al. (94) reported that the amplitude of mediolateral APA for gait initiation with the goal to clear an obstacle depended on whether participants had to strike the ground with a “rear foot strategy” (where the swing heel strikes first, RFS) or a “forefoot strategy” (where the toe strikes first, FFS). More specifically, it was found that the amplitude of the mediolateral APA was lower in the FFS than in the RFS condition. Striking the support surface with the toe first mechanically enlarged by a few centimeters the mediolateral base of support compared to striking with the heel first, thus making the control of stability in the final posture easier. As a consequence, the need to develop large mediolateral APA was lessened in the FFS condition as compared to the RFS condition to maintain stability. Because APA has an energy cost (46, 47, 95, 96), it was proposed that the CNS would reduce them in the former condition for economical purpose. Similar statement was proposed to explain why APA for upper limb task are depressed when the postural stability is made higher by an additional thoracic support (97). The mediolateral APA would be programmed according to the swing foot strike strategy and related-stability, thus revealing an interdependence between these two mechanisms. Globally, the current results thus add to the growing evidence that dynamic stability during gait initiation may share a principle of *homeostatic*-like regulation similar to most physiological variables, where separate mechanisms need to be coordinated to ensure stabilization of vital variables, here postural stability (31).

### Effects of Acute WBV on Postural Organization of Gait Initiation in the Elderly

In contrast to young adults, no significant effects of acute WBV application for any of the biomechanical variables recorded was observed in the elderly. This finding may a priori contrast with current literature that instead tends to show a beneficial effect of WBV application on postural control in static position and on various clinical locomotor tests [for review (98)]. Now, it is noteworthy that these latter studies used long-term WBV treatments and not single bout as in the present study. It is therefore not excluded that long-term WBV application would result in dynamic stability improvement in the elderly. The result that no increase in stance leg stiffness was observed in the WBV group post-treatment may possibly be ascribed to the progressive alteration with aging in the central structures (e.g., spinal motoneurons) (99, 100), somatosensory receptors of the lower limbs (e.g., spindles and mechanoreceptors of the

plantar foot sole) (101, 102), and/or sensory-motor pathways involved in the TVR reflex (103). This alteration would make elderly's stance leg stiffness less sensitive to acute WBV application than young adults. As a consequence, no motor adaptation e.g., in the form of larger APA or larger step width would be required to maintain stability since it was no further challenged by the WBV application. The current results however suggest that studies focusing on the effects of long-term WBV application in elderly should take into consideration the possibility that leg stiffness may be increased by this treatment. In case participants are not able to develop motor adaptation, instability may occur during locomotor tasks with increasing risks of falls.

## CONCLUSION

The present results showed that the effects of acute WBV application on the postural organization of gait initiation depend on the age of participants. In young healthy adults, a positive effect was observed on dynamic stability, while no effect was observed in the elderly. The positive effect was ascribed to an increase in the mediolateral APA following WBV application which overcompensated the potentially destabilizing effect of the increased stance leg stiffness. WBV application may thus be efficient to improve dynamic stability but at the condition that participants are able to develop adaptive changes in balance control mechanisms. In elderly, no anticipatory (nor corrective) postural adaptation was required since acute WBV application did not elicit change in the stance leg stiffness. Globally, the present finding is therefore in agreement with the hypothesis that balance control mechanisms are interdependent within the postural system, i.e., they may compensate each other in case one component (here the leg stiffness) is perturbed.

## DATA AVAILABILITY STATEMENT

The datasets generated for this study are available on request to the corresponding author.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by CIAMS Research Unit, Equipe d'Accueil (EA) 4532. The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

## AUTHOR CONTRIBUTIONS

AD, TV, MF, GL, LC, RA, FG, PF, and EY contributed with project creation, data collection, data analysis, drafted the manuscript, discussed the results, and participated in the revision of the manuscript.

## REFERENCES

- Mahieu NN, Witvrouw E, Van de Voorde D, Michielsens D, Arbyn V, Van den Broecke W. Improving strength and postural control in young skiers: whole-body vibration versus equivalent resistance training. *J Athl Train.* (2006) 41:286–93.
- Melnyk M, Kofler B, Faist M, Hodapp M, Gollhofer A. Effect of a whole-body vibration session on knee stability. *Int J Sports Med.* (2008) 29:839–44. doi: 10.1055/s-2008-1038405
- Ritzmann R, Kramer A, Bernhardt S, Gollhofer A. Whole body vibration training—improving balance control and muscle endurance. *PLoS ONE.* (2014) 9:e89905. doi: 10.1371/journal.pone.0089905
- Ritzmann R, Gollhofer A, Kramer A. The influence of vibration type, frequency, body position and additional load on the neuromuscular activity during whole body vibration. *Eur J Appl Physiol.* (2013) 113:1–11. doi: 10.1007/s00421-012-2402-0
- Torvinen S, Kannus P, Sievänen H, Järvinen TAH, Pasanen M, Kontulainen S, et al. Effect of four-month vertical whole body vibration on performance and balance. *Med Sci Sports Exerc.* (2002) 34:1523–8. doi: 10.1097/00005768-200209000-00020
- Torvinen S, Sievänen H, Järvinen TA, Pasanen M, Kontulainen S, Kannus P. Effect of 4-min vertical whole body vibration on muscle performance and body balance: a randomized cross-over study. *Int J Sports Med.* (2002) 23:374–9. doi: 10.1055/s-2002-33148
- Torvinen S, Kannus P, Sievänen H, Järvinen TA, Pasanen M, Kontulainen S, et al. Effect of 8-month vertical whole body vibration on bone, muscle performance, and body balance: a randomized controlled study. *J Bone Miner Res.* (2003) 18:876–84. doi: 10.1359/jbmr.2003.18.5.876
- Han S-W, Lee D-Y, Choi D-S, Han B, Kim J-S, Lee H-D. Asynchronous alterations of muscle force and tendon stiffness following 8 weeks of resistance exercise with whole-body vibration in older women. *J Aging Phys Act.* (2017) 25:287–94. doi: 10.1123/japa.2016-0149
- Rees SS, Murphy AJ, Watsford ML. Effects of whole body vibration on postural steadiness in an older population. *J Sci Med Sport.* (2009) 12:440–4. doi: 10.1016/j.jsams.2008.02.002
- Rees SS, Murphy AJ, Watsford ML. Effects of whole-body vibration exercise on lower-extremity muscle strength and power in an older population: a randomized clinical trial. *Phys Ther.* (2008) 88:462–70. doi: 10.2522/ptj.20070027
- Roelants M, Delecluse C, Verschueren SM. Whole-body-vibration training increases knee-extension strength and speed of movement in older women. *J Am Geriatr Soc.* (2004) 52:901–8. doi: 10.1111/j.1532-5415.2004.52256.x
- Rogan S, Radlinger L, Hilfiker R, Schmidtleicher D, de Bie RA, de Bruin ED. Feasibility and effects of applying stochastic resonance whole-body vibration on untrained elderly: a randomized crossover pilot study. *BMC Geriatr.* (2015) 15:25. doi: 10.1186/s12877-015-0021-4
- Yang F, King GA, Dillon L, Su X. Controlled whole-body vibration training reduces risk of falls among community-dwelling older adults. *J Biomech.* (2015) 48:3206–12. doi: 10.1016/j.jbiomech.2015.06.029
- Rogan S, Hilfiker R, Schenk A, Vogler A, Taeymans J. Effects of whole-body vibration with stochastic resonance on balance in persons with balance disability and falls history - a systematic review. *Res Sports Med.* (2014) 22:294–313. doi: 10.1080/15438627.2014.919504
- Piecha M, Juras G, Król P, Sobota G, Polak A, Bacik B. The effect of a short-term and long-term whole-body vibration in healthy men upon the postural stability. *PLoS ONE.* (2014) 9:e88295. doi: 10.1371/journal.pone.0088295
- Schlee G, Reckmann D, Milani TL. Whole body vibration training reduces plantar foot sensitivity but improves balance control of healthy subjects. *Neurosci Lett.* (2012) 506:70–3. doi: 10.1016/j.neulet.2011.10.051
- Hiroshige K, Mahbub MH, Harada N. Effects of whole-body vibration on postural balance and proprioception in healthy young and elderly subjects: a randomized cross-over study. *J Sports Med Phys Fitness.* (2014) 54:216–24.
- Orr R. The effect of whole body vibration exposure on balance and functional mobility in older adults: a systematic review and meta-analysis. *Maturitas.* (2015) 80:342–58. doi: 10.1016/j.maturitas.2014.12.020
- Rogan S, Taeymans J, Radlinger L, Naepflin S, Ruppen S, Bruehlhart Y, Hilfiker R. Effects of whole-body vibration on postural control in elderly: an update of a systematic review and meta-analysis. *Arch Gerontol Geriatr.* (2017) 73:95–112. doi: 10.1016/j.archger.2017.07.022
- Rogan S, Hilfiker R, Herren K, Radlinger L, de Bruin ED. Effects of whole-body vibration on postural control in elderly: a systematic review and meta-analysis. *BMC Geriatr.* (2011) 11:72. doi: 10.1186/1471-2318-11-72
- Blum L, Korner-Bitensky N. Usefulness of the Berg Balance Scale in stroke rehabilitation: a systematic review. *Phys Ther.* (2008) 88:559–66. doi: 10.2522/ptj.20070205
- Mancini M, Horak FB. The relevance of clinical balance assessment tools to differentiate balance deficits. *Eur J Phys Rehabil Med.* (2010) 46:239–48.
- Brenière Y, Cuong Do M, Bouisset S. Are dynamic phenomena prior to stepping essential to walking? *J Mot Behav.* (1987) 19:62–76. doi: 10.1080/00222895.1987.10735400
- Breniere Y, Do MC. When and how does steady state gait movement induced from upright posture begin? *J Biomech.* (1986) 19:1035–40. doi: 10.1016/0021-9290(86)90120-X
- Lepers R, Brenière Y. The role of anticipatory postural adjustments and gravity in gait initiation. *Exp Brain Res.* (1995) 107:118–24. doi: 10.1007/BF00228023
- Yiou E, Artico R, Teysseire CA, Labaune O, Fourcade P. Anticipatory postural control of stability during gait initiation over obstacles of different height and distance made under reaction-time and self-initiated instructions. *Front Hum Neurosci.* (2016) 10:449. doi: 10.3389/fnhum.2016.00449
- Caderby T, Yiou E, Peyrot N, Begon M, Dalleau G. Influence of gait speed on the control of mediolateral dynamic stability during gait initiation. *J Biomech.* (2014) 47:417–23. doi: 10.1016/j.jbiomech.2013.11.011
- Chong RKY, Chastan N, Welter M-L, Do M-C. Age-related changes in the center of mass velocity control during walking. *Neurosci Lett.* (2009) 458:23–7. doi: 10.1016/j.neulet.2009.04.022
- Honeine J-L, Schieppati M, Gagey O, Do M-C. The functional role of the triceps surae muscle during human locomotion. *PLoS ONE.* (2013) 8:52943. doi: 10.1371/journal.pone.0052943
- Honeine J-L, Schieppati M, Gagey O, Do M-C. By counteracting gravity, triceps surae sets both kinematics and kinetics of gait. *Physiol Rep.* (2014) 2:e00229. doi: 10.1002/phy2.229
- Yiou E, Caderby T, Delafontaine A, Fourcade P, Honeine J-L. Balance control during gait initiation: state-of-the-art and research perspectives. *World J Orthop.* (2017) 8:815–28. doi: 10.5312/wjo.v8.i11.815
- Barry E, Galvin R, Keogh C, Horgan F, Fahey T. Is the Timed Up and Go test a useful predictor of risk of falls in community dwelling older adults: a systematic review and meta-analysis. *BMC Geriatr.* (2014) 14:14. doi: 10.1186/1471-2318-14-14
- Topper AK, Maki BE, Holliday PJ. Are activity-based assessments of balance and gait in the elderly predictive of risk of falling and/or type of fall? *J Am Geriatr Soc.* (1993) 41:479–87. doi: 10.1111/j.1532-5415.1993.tb01881.x
- Delafontaine A, Gagey O, Colnaghi S, Do M-C, Honeine J-L. Rigid ankle foot orthosis deteriorates mediolateral balance control and vertical braking during gait initiation. *Front Hum Neurosci.* (2017) 11:214. doi: 10.3389/fnhum.2017.00214
- Honeine J-L, Schieppati M, Crisafulli O, Do M-C. The neuro-mechanical processes that underlie goal-directed medio-lateral APA during gait initiation. *Front Hum Neurosci.* (2016) 10:445. doi: 10.3389/fnhum.2016.00445
- Cau N, Cimolin V, Galli M, Precilios H, Tacchini E, Santovito C, et al. Center of pressure displacements during gait initiation in individuals with obesity. *J NeuroEng Rehabil.* (2014) 11:82. doi: 10.1186/1743-0003-11-82
- Zago M, Capodaglio P, Ferrario C, Tarabini M, Galli M. Whole-body vibration training in obese subjects: a systematic review. *PLoS ONE.* (2018) 13:e0202866. doi: 10.1371/journal.pone.0202866
- Cimolin V, Cau N, Galli M, Santovito C, Grugni G, Capodaglio P. Gait initiation and termination strategies in patients with Prader-Willi syndrome. *J Neuroeng Rehabil.* (2017) 14:44. doi: 10.1186/s12984-017-0257-7
- Sousa, Silva A, Santos R. Reliability of two methods for identifying the postural phase of gait initiation in healthy and poststroke subjects. *J Appl Biomech.* (2015) 31:349–56. doi: 10.1123/jab.2014-0222

40. Corsi C, Cimolin V, Capodaglio P, Condoluci C, Galli M. A biomechanical study of gait initiation in Down syndrome. *BMC Neurol.* (2019) 19:66. doi: 10.1186/s12883-019-1288-4
41. Chang H, Krebs DE. Dynamic balance control in elders: gait initiation assessment as a screening tool. *Arch Phys Med Rehabil.* (1999) 80:490–4. doi: 10.1016/S0003-9993(99)90187-9
42. Hass CJ, Waddell DE, Wolf SL, Juncos JL, Gregor RJ. Gait initiation in older adults with postural instability. *Clin Biomech.* (2008) 23:743–53. doi: 10.1016/j.clinbiomech.2008.02.012
43. Lee J, Webb G, Shortland AP, Edwards R, Wilce C, Jones GD. Reliability and feasibility of gait initiation centre-of-pressure excursions using a Wii® Balance Board in older adults at risk of falling. *Aging Clin Exp Res.* (2019) 31:257–63. doi: 10.1007/s40520-018-0945-6
44. Mbourou GA, Lajoie Y, Teasdale N. Step length variability at gait initiation in elderly fallers and non-fallers, and young adults. *Gerontology.* (2003) 49:21–6. doi: 10.1159/000066506
45. Crenna P, Frigo C. A motor programme for the initiation of forward-oriented movements in humans. *J Physiol.* (1991) 437:635–53. doi: 10.1113/jphysiol.1991.sp018616
46. Zettel JL, McLroy WE, Maki BE. Environmental constraints on foot trajectory reveal the capacity for modulation of anticipatory postural adjustments during rapid triggered stepping reactions. *Exp Brain Res.* (2002) 146:38–47. doi: 10.1007/s00221-002-1150-5
47. Zettel JL, McLroy WE, Maki BE. Can stabilizing features of rapid triggered stepping reactions be modulated to meet environmental constraints? *Exp Brain Res.* (2002) 145:297–308. doi: 10.1007/s00221-002-1083-z
48. Lyon IN, Day BL. Predictive control of body mass trajectory in a two-step sequence. *Exp Brain Res.* (2005) 161:193–200. doi: 10.1007/s00221-004-2058-z
49. Lyon IN, Day BL. Control of frontal plane body motion in human stepping. *Exp Brain Res.* (1997) 115:345–56. doi: 10.1007/PL00005703
50. Tisserand R, Robert T, Chabaud P, Bonnefoy M, Chèze L. Elderly fallers enhance dynamic stability through anticipatory postural adjustments during a choice stepping reaction time. *Front Hum Neurosci.* (2016) 10:613. doi: 10.3389/fnhum.2016.00613
51. You E, Caderby T, Hussein T. Adaptability of anticipatory postural adjustments associated with voluntary movement. *World J Orthop.* (2012) 3:75–86. doi: 10.5312/wjo.v3.i6.75
52. Roden-Reynolds DC, Walker MH, Wasserman CR, Dean JC. Hip proprioceptive feedback influences the control of mediolateral stability during human walking. *J Neurophysiol.* (2015) 114:2220–9. doi: 10.1152/jn.00551.2015
53. Ruget H, Blouin J, Coyle T, Mouchnino L. Modulation of proprioceptive inflow when initiating a step influences postural adjustments. *Exp Brain Res.* (2010) 201:297–305. doi: 10.1007/s00221-009-2035-7
54. Ruget H, Blouin J, Teasdale N, Mouchnino L. Can prepared anticipatory postural adjustments be updated by proprioception? *Neuroscience.* (2008) 155:640–8. doi: 10.1016/j.neuroscience.2008.06.021
55. Saradjian AH, Tremblay L, Perrier J, Blouin J, Mouchnino L. Cortical facilitation of proprioceptive inputs related to gravitational balance constraints during step preparation. *J Neurophysiol.* (2013) 110:397–407. doi: 10.1152/jn.00905.2012
56. Burke D, Hagbarth KE, Löfstedt L, Wallin BG. The responses of human muscle spindle endings to vibration during isometric contraction. *J Physiol.* (1976) 261:695–711. doi: 10.1113/jphysiol.1976.sp011581
57. Hagbarth KE, Eklund G. Tonic vibration reflexes (TVR) in spasticity. *Brain Res.* (1966) 2:201–3. doi: 10.1016/0006-8993(66)90029-1
58. Fowler BD, Palombo KTM, Feland JB, Blotter JD. Effects of whole-body vibration on flexibility and stiffness: a literature review. *Int J Exerc Sci.* (2019) 12:735–47.
59. Roschel H, Barroso R, Tricoli V, Batista MAB, Acquesta FM, Serrão JC, Ugrinowitsch C. Effects of strength training associated with whole-body vibration training on running economy and vertical stiffness. *J Strength Cond Res.* (2015) 29:2215–20. doi: 10.1519/JSC.0000000000000857
60. Wang P, Yang X, Yang Y, Yang L, Zhou Y, Liu C, et al. Effects of whole body vibration on pain, stiffness and physical functions in patients with knee osteoarthritis: a systematic review and meta-analysis. *Clin Rehabil.* (2015) 29:939–51. doi: 10.1177/0269215514564895
61. Ness LL, Field-Fote EC. Whole-body vibration improves walking function in individuals with spinal cord injury: a pilot study. *Gait Posture.* (2009) 30:436–40. doi: 10.1016/j.gaitpost.2009.06.016
62. van Nes IJW, Latour H, Schils F, Meijer R, van Kuijk A, Geurts ACH. Long-term effects of 6-week whole-body vibration on balance recovery and activities of daily living in the postacute phase of stroke: a randomized, controlled trial. *Stroke.* (2006) 37:2331–5. doi: 10.1161/01.STR.0000236494.62957.f3
63. van Nes IJW, Geurts ACH, Hendricks HT, Duysens J. Short-term effects of whole-body vibration on postural control in unilateral chronic stroke patients: preliminary evidence. *Am J Phys Med Rehabil.* (2004) 83:867–73. doi: 10.1097/01.PHM.0000140801.23135.09
64. Farley CT, Morgenroth DC. Leg stiffness primarily depends on ankle stiffness during human hopping. *J Biomech.* (1999) 32:267–73. doi: 10.1016/S0021-9290(98)00170-5
65. Morasso PG, Schieppati M. Can muscle stiffness alone stabilize upright standing? *J Neurophysiol.* (1999) 82:1622–6. doi: 10.1152/jn.1999.82.3.1622
66. Brenière Y, Do MC. Control of gait initiation. *J Mot Behav.* (1991) 23:235–40. doi: 10.1080/00222895.1991.9942034
67. Caderby T, You E, Peyrot N, Bonazzi B, Dalleau G. Detection of swing heel-off event in gait initiation using force-plate data. *Gait Posture.* (2013) 37:463–6. doi: 10.1016/j.gaitpost.2012.08.011
68. Delafontaine A, Honeine J-L, Do M-C, Gagey O, Chong RK. Comparative gait initiation kinematics between simulated unilateral and bilateral ankle hypomobility: does bilateral constraint improve speed performance? *Neurosci Lett.* (2015) 603:55–9. doi: 10.1016/j.neulet.2015.07.016
69. You E, Teysède C, Artico R, Fourcade P. Comparison of base of support size during gait initiation using force-plate and motion-capture system: a Bland and Altman analysis. *J Biomech.* (2016) 49:4168–72. doi: 10.1016/j.jbiomech.2016.11.008
70. Hof AL, Gazendam MGJ, Sinke WE. The condition for dynamic stability. *J Biomech.* (2005) 38:1–8. doi: 10.1016/j.jbiomech.2004.03.025
71. Caderby T, You E, Peyrot N, de Viviés X, Bonazzi B, Dalleau G. Effects of changing body weight distribution on mediolateral stability control during gait initiation. *Front Hum Neurosci.* (2017) 11:127. doi: 10.3389/fnhum.2017.00127
72. You E, Fourcade P, Artico R, Caderby T. Influence of temporal pressure constraint on the biomechanical organization of gait initiation made with or without an obstacle to clear. *Exp Brain Res.* (2016) 234:1363–75. doi: 10.1007/s00221-015-4319-4
73. Winter DA, Patla AE, Frank JS. Assessment of balance control in humans. *Med Prog Technol.* (1990) 16:31–51.
74. Ditcher S, You E, Delafontaine A, Hamaoui A. Short-term effects of thoracic spine manipulation on the biomechanical organisation of gait initiation: a randomized pilot study. *Front Hum Neurosci.* (2017) 11:343. doi: 10.3389/fnhum.2017.00343
75. Do MC, Bussel B, Brenière Y. Influence of plantar cutaneous afferents on early compensatory reactions to forward fall. *Exp Brain Res.* (1990) 79:319–24. doi: 10.1007/BF00608241
76. Mouchnino L, Blouin J. When standing on a moving support, cutaneous inputs provide sufficient information to plan the anticipatory postural adjustments for gait initiation. *PLoS ONE.* (2013) 8:e55081. doi: 10.1371/journal.pone.0055081
77. Thoumie P, Do MC. Changes in motor activity and biomechanics during balance recovery following cutaneous and muscular deafferentation. *Exp Brain Res.* (1996) 110:289–97. doi: 10.1007/BF00228559
78. You E, Ditcher S, Le Bozec S. Biomechanical reorganisation of stepping initiation during acute dorsiflexor fatigue. *J Electromyogr Kinesiol.* (2011) 21:727–33. doi: 10.1016/j.jelekin.2011.04.008
79. Alfuth M, Beiring A, Klein D, Rosenbaum D. Acute effects of whole body vibration on foot sole sensitivity and plantar pressures during gait initiation. *J Foot Ankle Res.* (2012) 5:O24. doi: 10.1186/1757-1146-5-S1-O24
80. Pollock RD, Provan S, Martin FC, Newham DJ. The effects of whole body vibration on balance, joint position sense and cutaneous sensation. *Eur J Appl Physiol.* (2011) 111:3069–77. doi: 10.1007/s00421-011-1943-y
81. Pollock RD, Provan S, Martin FC, Newham DJ. Whole body vibration; short-term effects on balance, proprioception and cutaneous sensation

- in healthy young adults. *Proc of The Phys Soc.* (2010) 19:3069–77. doi: 10.3138/ptc.2014-77
82. Hannah R, Minshull C, Folland JP. Whole-body vibration does not influence knee joint neuromuscular function or proprioception. *Scand J Med Sci Sports.* (2013) 23:96–104. doi: 10.1111/j.1600-0838.2011.01361.x
  83. Lee TY, Chow DHK. Effects of whole body vibration on spinal proprioception in normal individuals. *Conf Proc IEEE Eng Med Biol Soc.* (2013) 2013:4989–92. doi: 10.1109/EMBC.2013.6610668
  84. Li L, Lamis F, Wilson SE. Whole-body vibration alters proprioception in the trunk. *Int J Ind Ergon.* (2008) 38:792–800. doi: 10.1016/j.ergon.2007.10.010
  85. Lin SJ, Yang WC. Effect of plantar desensitization on postural adjustments prior to step initiation. *Gait Posture.* (2011) 34:451–6. doi: 10.1016/j.gaitpost.2011.06.016
  86. Madeleine P, Voigt M, Arendt-Nielsen L. Reorganisation of human step initiation during acute experimental muscle pain. *Gait Posture.* (1999) 10:240–7. doi: 10.1016/S0966-6362(99)00036-3
  87. Rieder F, Wiesinger H-P, Kösters A, Müller E, Seynnes OR. Immediate effects of whole body vibration on patellar tendon properties and knee extension torque. *Eur J Appl Physiol.* (2016) 116:553–61. doi: 10.1007/s00421-015-3316-4
  88. Siu PM, Tam BT, Chow DH, Guo J-Y, Huang Y-P, Zheng Y-P, Wong SH. Immediate effects of 2 different whole-body vibration frequencies on muscle peak torque and stiffness. *Arch Phys Med Rehabil.* (2010) 91:1608–15. doi: 10.1016/j.apmr.2010.07.214
  89. Cronin JB, Oliver M, McNair PJ. Muscle stiffness and injury effects of whole body vibration. *Phys Ther Sport.* (2004) 5:68–74. doi: 10.1016/S1466-853X(04)00020-3
  90. Colson SS, Petit P-D. Lower limbs power and stiffness after whole-body vibration. *Int J Sports Med.* (2013) 34:318–23. doi: 10.1055/s-0032-1311596
  91. Sandhu E, Miles JD, Dahners LE, Keller BV, Weinhold PS. Whole body vibration increases area and stiffness of the flexor carpi ulnaris tendon in the rat. *J Biomech.* (2011) 44:1189–91. doi: 10.1016/j.jbiomech.2011.02.017
  92. Delafontaine A, Fourcade P, Honeine JL, Ditcher S, Yiu E. Postural adaptations to unilateral knee joint hypomobility induced by orthosis wear during gait initiation. *Sci Rep.* (2018) 8:830. doi: 10.1038/s41598-018-19151-1
  93. Reimann H, Fettrow TD, Thompson ED, Agada P, McFadyen BJ, Jeka JJ. Complementary mechanisms for upright balance during walking. *PLoS ONE.* (2017) 12:e0172215. doi: 10.1371/journal.pone.0172215
  94. Artico R, Fourcade P, Teyssedre CA, Delafontaine A, Yiu E. Influence of swing foot strike pattern on balance control mechanisms during gait initiation over an obstacle to clear. In: *Progress in Motor Control.* Amsterdam (2019) Available online at: <https://10times.com/pmc-netherlands>
  95. Aruin AS, Forrest WR, Latash ML. Anticipatory postural adjustments in conditions of postural instability. *Electroencephalogr Clin Neurophysiol.* (1998) 109:350–9. doi: 10.1016/S0924-980X(98)00029-0
  96. Nouillot P, Bouisset S, Do MC. Do fast voluntary movements necessitate anticipatory postural adjustments even if equilibrium is unstable? *Neurosci Lett.* (1992) 147:1–4. doi: 10.1016/0304-3940(92)90760-5
  97. Cordo PJ, Nashner LM. Properties of postural adjustments associated with rapid arm movements. *J Neurophysiol.* (1982) 47:287–302. doi: 10.1152/jn.1982.47.2.287
  98. Fischer M, Vialleron T, Laffaye G, Fourcade P, Hussein T, Chèze L, et al. Long-term effects of whole-body vibration on human gait: a systematic review and meta-analysis. *Front Neurol.* (2019) 10:627. doi: 10.3389/fneur.2019.00627
  99. Varma VR, Hausdorff JM, Studenski SA, Rosano C, Camicioli R, Alexander NB, et al. Aging, the central nervous system, and mobility in older adults: interventions. *J Gerontol A Biol Sci Med Sci.* (2016) 71:1451–8. doi: 10.1093/gerona/glw080
  100. Edström E, Altun M, Bergman E, Johnson H, Kullberg S, Ramírez-León V, et al. Factors contributing to neuromuscular impairment and sarcopenia during aging. *Physiol Behav.* (2007) 92:129–35. doi: 10.1016/j.physbeh.2007.05.040
  101. Liu J-X, Eriksson P-O, Thornell L-E, Pedrosa-Domellöf F. Fiber content and myosin heavy chain composition of muscle spindles in aged human biceps brachii. *J Histochem Cytochem.* (2005) 53:445–54. doi: 10.1369/jhc.4A6257.2005
  102. Peters RM, McKeown MD, Carpenter MG, Inglis JT. Losing touch: age-related changes in plantar skin sensitivity, lower limb cutaneous reflex strength, and postural stability in older adults. *J Neurophysiol.* (2016) 116:1848–58. doi: 10.1152/jn.00339.2016
  103. Shaffer SW, Harrison AL. Aging of the somatosensory system: a translational perspective. *Phys Ther.* (2007) 87:193–207. doi: 10.2522/ptj.20060083

**Conflict of Interest:** The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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## **3.2. Effets des étirements musculaires sur la marche et l'initiation de la marche**

### **3.2.1. Effets à long terme des étirements musculaires sur la marche : revue systématique de la littérature et méta-analyse**

Cette étude a donné lieu à la publication ci-dessous :

**Vialleron T**, Delafontaine A, Ditcharles S, Fourcade P, Yiou E (2020). Effects of stretching exercises on human gait: a systematic review and meta-analysis. *F1000Res* 13; 9: 984  
Doi: 10.12688/f1000research.25570.2.



## SYSTEMATIC REVIEW

**REVISED** Effects of stretching exercises on human gait: a systematic review and meta-analysis [version 2; peer review: 2 approved, 1 approved with reservations]

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**Abstract**

**Background:** Stretching is commonly used in physical therapy as a rehabilitation tool to improve range of motion and motor function. However, is stretching an efficient method to improve gait, and if so, for which patient category?

**Methods:** A systematic review of randomized and non-randomized controlled trials with meta-analysis was conducted using relevant databases. Every patient category and every type of stretching programs were included without multicomponent programs. Data were meta-analysed where possible. Estimates of effect sizes (reported as standard mean difference (SMD)) with their respective 95% confidence interval (95% CI) were reported for each outcome. The PEDro scale was used for the quality assessment.

**Results:** Twelve studies were included in the analysis. Stretching improved gait performance as assessed by walking speed and stride length only in a study with a frail elderly population, with small effect sizes (both SMD= 0.49; 95% CI: 0.03, 0.96; PEDro score: 3/10). The total distance and the continuous walking distance of the six-minute walking test were also improved only in a study in an elderly population who had symptomatic peripheral artery disease, with large effect sizes (SMD= 1.56; 95% CI: 0.66, 2.45 and SMD= 3.05; 95% CI: 1.86, 4.23, respectively; PEDro score: 5/10). The results were conflicting in healthy older adults or no benefit was found for most of the performance, spatiotemporal, kinetic and angular related variables. Only one study (PEDro score: 6/10) showed improvements in stance phase duration (SMD=-1.92; 95% CI: -3.04, -0.81), swing phase duration (SMD=1.92; 95% CI: 0.81, 3.04), double support phase duration (SMD= -1.69; 95% CI: -2.76, -0.62) and step length (SMD=1.37; 95% CI: 0.36, 2.38) with large effect sizes.

**Conclusions:** There is no strong evidence supporting the beneficial

**Open Peer Review**

Reviewer Status

	Invited Reviewers		
	1	2	3
<b>version 2</b> (revision) 30 Oct 2020	 report		 report
<b>version 1</b> 13 Aug 2020	 report	 report	

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effect of using stretching to improve gait. Further randomized controlled trials are needed to understand the impact of stretching on human gait.

**Keywords**

stretching, gait, performance, balance, physical therapy

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Any reports and responses or comments on the article can be found at the end of the article.

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**REVISED Amendments from Version 1**

We have added a whole paragraph at the beginning of the introduction section to specify how gait can be related to the different variables seen in the review.

No article in children with cerebral palsy fitted our inclusion criteria during the systematic search of the literature. We have added a whole paragraph in a specific section "limitations of the study" for explain it.

We have added some precisions in the discussion section (healthy older adults paragraph) to specify how we have limited the risk of bias.

We have added some precisions in the discussion section (young adults paragraph) to specify why stretching is less interesting to improve gait parameters in young healthy adults.

**Any further responses from the reviewers can be found at the end of the article**

**Introduction**

Gait is the medical term used to describe the human whole body movement of walking<sup>1</sup>. Gait involves internal and external forces that act on the body to move the center of mass (COM) across a given distance<sup>2</sup>. It depends on many biomechanical features that can be observed during gait analysis such as center of mass shift, joint range of motion (ROM), forces, muscle activity, joint moments, and joint powers<sup>3</sup>. Spatiotemporal features (e.g. velocity, step length, stride length, step width, step variability) and kinematics parameters (ROM) can be observed subjectively with functional evaluations by clinicians (e.g. the Tinetti test<sup>4</sup> or the timed up and go test<sup>5</sup>), but, it can be further objectified with biomechanical analysis in a laboratory<sup>2</sup>. Kinetics variables (the forces that cause the body to move) must be collected in a laboratory environment with force plates (e.g. 6–9 for recent studies that used this technic).

Gait is a highly complex motor skill that is classically considered as an integrative measure and a predictor of health in older adults (e.g. 10; cf. 11 and also 12 for recent research topics on this matter). The loss of gait or its alteration with pathological conditions are known to be related to mortality, especially in the elderly (e.g. 13,14), stressing the importance of addressing gait disorders in physiotherapy. Gait requires body propulsion and balance control for safe progression, two "subtasks" that require the coordination of multiple skeletal muscles and the integration of sensory information arising from the vestibular, visual and somatosensory systems<sup>15–17</sup>. As such, gait may expose populations with sensory or motor deficits to the risk of falling with serious consequences for health and autonomy. For these reasons, improving gait is a major aim in rehabilitation for most neurological/orthopaedic disorders, such as stroke or Parkinson's disease, and for frail older adults. Various therapeutic methods have been used to improve gait, such as resistance training<sup>18</sup>, endurance training<sup>19</sup>, balance training<sup>20</sup>, whole body vibrations (for a complete review, see Fischer *et al.*, 2019<sup>21</sup>), multi-component exercise programs<sup>22</sup> and stretching<sup>23</sup>.

The successful completion of numerous daily life activities is conditioned by the ability to move efficiently through a sufficient ROM<sup>24</sup>. Recent studies on gait initiation<sup>25–27</sup> and

seat-to-stand task<sup>28,29</sup> showed that the experimental restriction of postural chain ROM induced by orthosis wear in young healthy adults led to instability and lower motor performance. It is well established that ROM significantly decreases with aging<sup>30–35</sup> and more generally with reduced functional demand (e.g. sedentarity, immobilization, disease etc.)<sup>24</sup>. Consequently, stretching has become an important part of many sport and rehabilitation programs to maintain or improve ROM, reduce stiffness and promote physical activity. This method has been applied in older adults<sup>36,37</sup>, patients with stroke<sup>38</sup>, Parkinson's disease<sup>39</sup>, multiple sclerosis<sup>40</sup>, plantar fasciitis<sup>41</sup> and spastic paraplegia<sup>42</sup>, for example. In sport programs, the influence of stretching on motor performance remains an issue of debate, although recent reviews conclude that maximal muscle performance (e.g. force, power, jump height, reaction time, etc.) is impaired primarily immediately after long durations of stretch (>90 seconds)<sup>43,44</sup>. To date, no review has collected results on the relationship between stretching and locomotor performance in rehabilitation programs.

Hence, the purpose of this article is to analyse the effects of a stretching program on gait in each patient category by means of a systematic literature review and meta-analysis, comparing the gait outcomes of the intervention groups with the control groups. It will contribute to provide evidence-based practice from scientific data in order to integrate stretching in rehabilitation programs in a reasoned manner.

**Methods****Design and literature screening**

The Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) methodology was employed in this systematic review<sup>45</sup>. A completed PRISMA checklist was submitted to an online repository (*Reporting guidelines*).

PubMed, Science Direct, Springer and Sage databases were used for a comprehensive systematic literature search for articles published prior to 28 April 2020 with no time limit. In addition, a manual search was conducted using the reference list of selected studies. The keywords used for the search strategy in PubMed were: "stretching" AND (gait OR walk). We included only articles published in English or French.

The selection procedure was conducted by two experts in rehabilitation (TV and AD). Disagreements were discussed with a third expert in a group until a mutual consensus was reached. First, a review was performed on all available titles obtained from the literature search with the selected keywords. All relevant or potentially relevant titles were included in the subsequent phase. Then, the abstracts were reviewed with all relevant or potential articles included in the following phase. Finally, full-text articles were reviewed to ensure that only relevant studies were included. In the same way, reference lists of all included articles were reviewed to possibly include articles through cross-referencing.

**Inclusion and exclusion criteria**

We included randomized controlled trials (RCT) and controlled clinical trials (CCT) published in peer-reviewed journals that aimed to explore the effects of stretching on gait parameters.

We included all categories of subjects, all stretching techniques and different durations of treatment since standardized protocols are lacking in the purpose of the present study. Gait could be evaluated by functional tests, electromyographic (EMG) or biomechanical analysis. The following exclusion criteria were used: lack of gait assessment, non-application of muscle stretching, multimodal exercise programs, no control group, case report and review.

#### Data extraction and main measurements examined

Data were extracted from the selected articles by one of the authors (TV). The extracted data were checked by another author (AD) and disagreements were resolved with a third (EY).

The following data were extracted for each selected article: (1) the names of the authors and the date of publication; (2) the number of subjects involved in the experiment with their characteristics and breakdown in each group; (3) stretching training details (in the following order: number of participants, stretching technique, muscle groups stretched, number of sets, duration of stretch, frequency, protocol duration); (4) control group details; and (5) the main outcomes related to gait with the main results. When information could not be provided, it was indicated by a “?”.

#### Quality and risk of bias assessment

The PEDro scale was used to assess the risk of bias, and thus the methodological quality of the selected studies<sup>46</sup>. This scale was chosen for its ability to provide an overview of the external (criterion 1), internal (criteria 2–9) and statistical (criteria 9 and 10) validity of clinical trials. The scale is divided into 11 criteria, but the first is not calculated in the total score. The output of each criterion could be either “yes” (y), “no” (n) or “do not know” (?). A “y” was given a score of one point, while an “n” or “?” was assigned zero points. Studies with a total score of 5–10/10 ( $\geq 50\%$ ) were considered to be of high quality, and scores of 0–4/10 ( $<50\%$ ) of low quality<sup>47</sup>. Two evaluators independently assessed the quality of the included studies. In the event of disagreement, a group discussion was held with a third expert to reach a consensus.

#### Statistical analysis

Estimates of effect sizes (comparing the intervention group and the control group) accompanied with a measure of statistical uncertainty (95% confidence interval [95% CI]) were calculated for each outcome when sufficient data were reported. Estimates of effect sizes were reported by standard mean difference (SMD) and their respective 95% CI. In this way, the magnitude of the overall effect can be quantified as trivial ( $<0.2$ ), small (0.2–0.49), moderate (0.5–0.79) or large ( $\geq 0.8$ )<sup>48,49</sup>. When data were lacking to calculate estimates of effect sizes, exact *p* values were reported.

When at least two studies used the same outcome, meta-analysis was performed, comparing the intervention groups with the control groups. When outcomes were identified in only one study, no meta-analysis could be performed but the effect of intervention was still calculated, reporting the estimate of effect size and its 95% confidence interval. Statistical analysis and figures (i.e. forest plot to facilitate the visualization of

values) were produced using a random-effect model in Review Manager software (RevMan, v 5.3, Cochrane Collaboration, Oxford UK). A random-effect model was used to take into account heterogeneity between study effects. Statistical heterogeneity was calculated using the  $I^2$  and Cochrane *Q* statistic tests<sup>48</sup>. Statistical significance was set at  $p < 0.05$ .

#### Level of evidence

The strength of evidence of primary outcomes was established as described by Van Tulder *et al.* 2003<sup>50</sup> based on effect size estimates with a measure of statistical uncertainty (SMD; 95% CI), statistical heterogeneity ( $I^2$ ) when applicable (multiple studies) and risk of bias (PEDro scale). The level of evidence was considered strong with consistent findings among multiple high-quality RCT (at least two RCT with a PEDro score  $\geq 5/10$  that were statistically homogenous:  $I^2 p \geq 0.05$ ). The level of evidence was considered moderate with consistent findings among multiple low-quality RCT and/or CCT (two trials with a PEDro score  $< 5/10$  that were statistically homogenous) and/or one high quality RCT. The level of evidence was considered limited when only one low quality RCT and/or CCT was identified. The level of evidence was conflicting when there was inconsistency among multiple trials ( $I^2 p < 0.05$ ).

## Results

### Included studies

A total of 821 titles were screened in the first search stage, one more was included through cross-referencing, and 671 were excluded because they did not concern our research question. Following exclusion, 150 studies were considered for an abstract review. A further 105 were excluded in this second stage because they did not meet the inclusion criteria. Finally, 45 full-text articles were assessed for eligibility with 33 not accepted (Figure 1).

Thus, 12 articles were ultimately included in this systematic review. Six studies evaluated the effects of stretching in healthy older adults<sup>23,51–55</sup>, one in a frail elderly population<sup>56</sup>, one study in an elderly population with stable symptomatic peripheral artery disease<sup>57</sup>, one in stroke patients<sup>58</sup>, one study in adults with limited ankle ROM associated with a history of lower limb overuses injury<sup>59</sup>, one study in healthy adults with limited ankle dorsiflexion<sup>60</sup> and one in healthy young adults<sup>61</sup>. A summary of the studies selected is provided in Table 1, and their quality assessment is reported in Table 2. The results in different patient categories are reported below.

### Results in different patient categories

#### Healthy older adults

##### Description of the studies and quality assessment

Six studies examined the effects of stretching on healthy elderly subjects<sup>23,51–55</sup>. Regarding the characteristics of the subjects, the average sample size was  $46.6 \pm 33.9$  subjects (ranging from 19<sup>23</sup> to 96 subjects<sup>53</sup>) and the mean age was  $70.1 \pm 3.6$  years (ranging from 65.40<sup>52</sup> to 75.30 years<sup>23</sup>). Regarding the characteristics of the training programs, the average training duration was  $8.6 \pm 2.7$  weeks (ranging from 4<sup>52</sup> to 12 weeks<sup>54</sup>), with an average frequency of  $8.3 \pm 6.2$  sessions per week (ranging from 2<sup>54</sup> to 14 sessions<sup>51,53,55</sup>). The average number of sets per



## PRISMA 2009 Flow Diagram

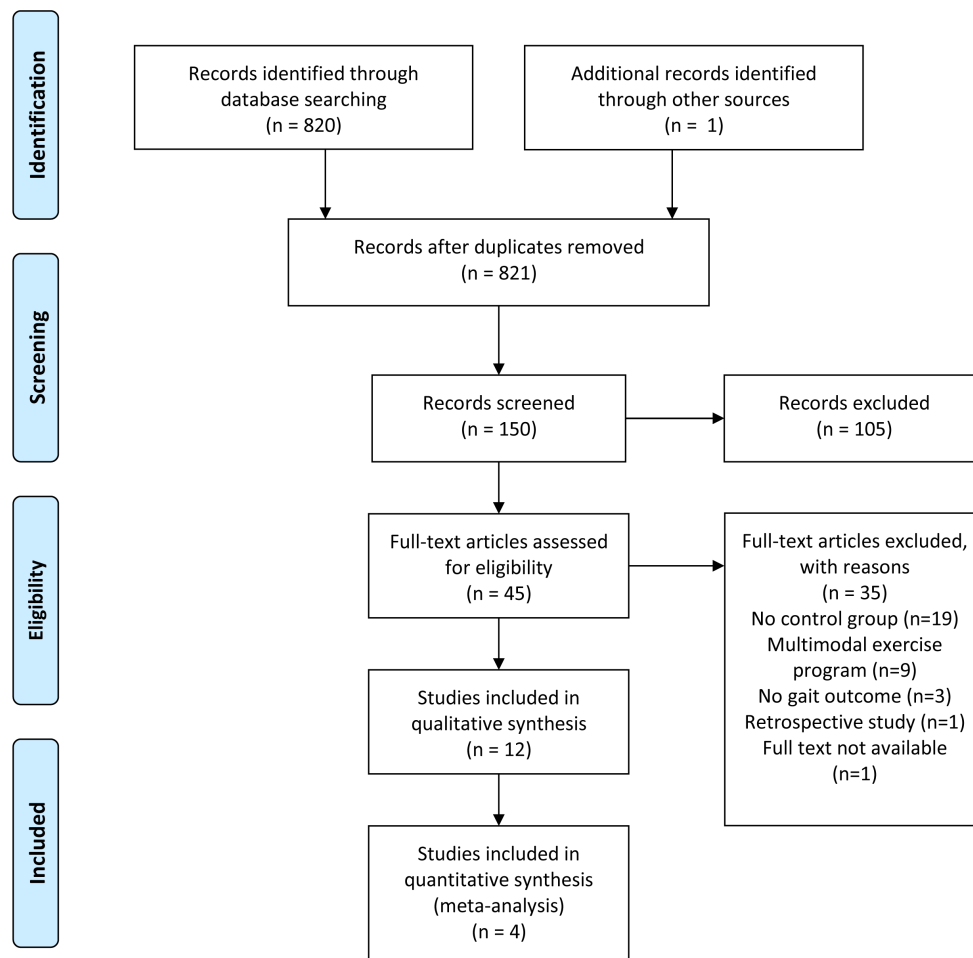


Figure 1. PRISMA flow chart of study selection process.

session was  $4.5 \pm 2.8$  sets (ranging from 2<sup>55</sup> to 10 sets<sup>23</sup>), with an average stretching time of  $45.0 \pm 18.9$  seconds (ranging from 15<sup>23</sup> to 60 seconds<sup>52,54,55</sup>). Static stretching was provided in all studies. The muscle groups stretched were the hip flexors<sup>51-55</sup>, ankle plantar flexors<sup>23,51,52,54</sup>, ankle dorsiflexors<sup>54</sup>, hip extensors<sup>52</sup>, knee extensors and flexors<sup>54</sup>. There was great heterogeneity in gait outcomes. Angular variables during gait included peak hip extension<sup>51,53,55</sup>, ankle plantar flexion during gait<sup>53</sup>, ankle ROM during gait<sup>52</sup>, anterior pelvis tilt<sup>52,55</sup>, knee ROM, pelvic rotation,

lateral pelvic tilt and hip ROM<sup>52</sup>. Spatiotemporal variables were: gait speed<sup>51,52,55</sup>, stance and swing duration, double support phases, step length<sup>52</sup> and stride length<sup>52,55</sup>. Kinetic variables were hip torque and ankle plantar flexion power<sup>53</sup>. Finally, two functional tests were used: the 10-meter walk test (10MWT)<sup>23</sup> and the 6-minute walk test (6MWT)<sup>54</sup>. Regarding the quality of the studies, the average PEDro score was  $4.6 \pm 1.6$  and one study was identified as a non-randomized trial<sup>54</sup>. The range of score varied from 3<sup>54,55</sup> to 7<sup>23</sup>.

**Table 1. Summary of the included studies.**

Studies	Population	Stretching group	Control group	Outcomes and main results
Kerrigan <i>et al.</i> , 2003 <sup>53</sup>	96 healthy older adults, ? (≥65 years)	n= 47, static stretching, hip flexors, 4 sets, 30 seconds, twice daily, 10 weeks	n=49, static shoulder deltoid-stretching, same protocol	No significant difference between groups for hip extension (SMD= 0.22; 95% CI: -0.18, 0.62), hip torque (SMD= 0.35; 95% CI: -0.06, 0.75), anterior pelvic tilt (SMD= -0.35; 95% CI : -0.76, 0.05), ankle plantar flexion ROM (SMD= -0.05, 95% CI : -0.45, 0.35), ankle plantar flexion power (SMD= 0.00; 95% CI : -0.40, 0.40), hip extension (SMD= 0.22; 95% CI : -0.19, 0.62) and hip torque (SMD= 0.35; 95% CI: -0.06, 0.75)
Gajdosik <i>et al.</i> , 2005 <sup>23</sup>	19 community dwelling older women, ? (65–89 years)	n=10, static stretching, ankle plantar flexors, 10 sets, 15 seconds, 3 times per week, 8 weeks	n=9, no exercise.	No significant difference between groups for 10MWT (SMD= -0.76; 95% CI: -1.70, 0.18)
Christiansen, 2008 <sup>51</sup>	40 healthy older adults, 72.10±4.70 years	N= 20, static stretching, hip flexors, 3 sets, 45 seconds, twice daily, 8 weeks	n=20, maintain their current level of physical activity	No significant difference between groups for gait speed (SMD= -0.32; 95% CI: -0.97, 0.33), hip extension (SMD= 0.22; 95% CI: -0.43, 0.86), stride length (SMD= - 0.14; 95% CI : -0.79, 0.50), ankle dorsiflexion (SMD= 0.29; 95% CI : -0.36, 0.94)
Cristopoliski <i>et al.</i> , 2009 <sup>52</sup>	20 healthy elderly women, 65.90±4.20 years	n=12, static stretching, hip flexors and extensors, ankle plantar flexors, 4 sets, 60 seconds, 3 sessions per week, 12 sessions	n=8, no specific activity in this period	Significant improvement in favor of stretching group for gait speed (SMD= 1.32; 95% CI : 0.32, 2.32), anterior pelvic tilt (SMD= -2.52; 95% CI : -3.77, -1.27), stand phase duration (SMD= -1.92; 95% CI: -3.04, -0.81), swing phase duration (SMD= 1.92; 95% CI : 0.81, 3.04), double support phase duration (SMD= -1.69; 95% CI: -2.76, -0.62), step length (SMD= 1.37; 95% CI : 0.36, 2.38) and pelvic rotation (SMD= 1.37; 95% CI : 0.36, 2.38) No significant difference between groups for cycle duration (SMD= - 0.24; 95% CI : -1.14, 0.66), heel-contact velocity (SMD= -0.46; 95% CI : -1.37, 0.45), toe clearance (SMD= 0.91; 95% CI : -0.04, 1.86), lateral pelvic tilt (SMD= 0.93; 95% CI : -0.02, 1.88) and knee ROM (SMD= 0.23; 95% CI : -0.67, 1.12)
Watt <i>et al.</i> , 2011 <sup>55</sup>	82 healthy elderly subjects, 72,6±6 years	n= 43, static stretching, hip flexors, 2 sets, 60 seconds, 2 sessions daily stretching, 10 weeks	N= 39, shoulder abductor static stretching, same protocol	Significant improvement in favor of stretching group for gait speed (SMD= 0.47; 95% CI: 0.03, 0.91) No significant difference between groups for hip extension (SMD= 0.18; 95% CI: -0.25, 0.62), anterior pelvic tilt (SMD=0.07; 95% CI : -0.36, 0.51), stride length (SMD= 0.54; 95% CI : -0.01, 1.08)
Locks <i>et al.</i> , 2012 <sup>54</sup>	23 healthy older individuals, 67.5±2.12 years	n=10, static stretching, knee extensors, ankle dorsiflexor, knee flexors, ankle plantar flexors, 4 sets, 60 seconds, twice a week ,12 weeks	n=13 a one-hour seminar on healthy living every four weeks and did not perform any physical or therapeutic exercise.	No significant difference between groups for 6MWT (SMD= -0.04; 95% CI : -0.86, 0.79)
Watt <i>et al.</i> , 2011 <sup>56</sup>	74 frail elderly subjects, 77.00±8.00 years	n=33, static stretching, hip flexors, 2 sets, 60 seconds, 2 sessions per day, 10 weeks	n=41, shoulder abductor stretching program, same protocol	No significant difference between groups in peak hip extension, (SMD= 0.22; 95% CI: -0.24, 0.68), anterior pelvic tilt (SMD= -0.05; 95% CI: -0.51, 0.41) and cadence (SMD= 0.13; 95% CI: -0.33, 0.59) Significant improvements in favor of the stretching group in walking speed and stride length (both SMD= 0.49; 95% CI: 0.03, 0.96)

Studies	Population	Stretching group	Control group	Outcomes and main results
Hotta <i>et al.</i> , 2019 <sup>57</sup>	13 elderly patients with symptomatic peripheral artery disease, ?	n= 13, static stretching, ankle plantar flexor stretching, 1 set, 30 minutes, 5 sessions per week, 4 weeks	n= 13, no stretching intervention (cross-over intervention)	Significant improvements in favor of the stretching group for both total walking distance and continuous walking distance with large effect sizes (SMD= 1.56; 95% CI: 0.66, 2.45 and SMD= 3.05; 95% CI: 1.86, 4.23 respectively)
Kim <i>et al.</i> , 2013 <sup>58</sup>	24 patients with stroke, 53.30±3.16 years	n=12, static stretching, ankle plantar flexors, 1 set, 20 minutes, 4 times a week, 4 times a week, 6 weeks	n= 12, conventional physical therapy as in in the stretching group	No significant difference between groups in sway of the center of pressure (SMD=0.75; 95% CI: -0.09, 1.58)
Johanson <i>et al.</i> , 2006 <sup>59</sup>	19 adults with limited passive ankle-dorsiflexion ROM (less than 8 degrees) and a history of lower limb overuse injury, 30.30±9.80 years	n=11, static stretching, ankle plantar flexors, 5 sets, 30 seconds, 2 times daily, 3 weeks	n= 8, continue all of their usual activities	No significant difference between groups in ankle dorsiflexion during gait in both right and left ankle (SMD= 0.50; 95% CI: -0.42, 1.43 and SMD= 0.41; 95% CI: -0.52, 1.33 respectively) and for time-to-heel-off during the stance phase of gait in both right and left ankle (SMD= -0.50; 95% CI: -1.43, 0.43 and SMD= -0.48; 95% CI: -1.41, 0.45 respectively)
Johanson <i>et al.</i> , 2009 <sup>60</sup>	16 healthy adults with limited passive ankle-dorsiflexion ROM (less than 5 degrees), 27.40±8.20 years	n=8, static stretching, ankle plantar flexors, 4 sets, 30 seconds, 2 times daily, 3 weeks	n=8, no physical activity or stretching programs involving the lower extremities for 3 weeks	No significant difference between groups in ankle dorsiflexion (SMD= 0.53; 95% CI: -0.48, 1.53), maximum knee extension (SMD= -0.07; 95% CI: -1.05, 0.91), medial and lateral gastrocnemius activities (SMD= 0.37; 95% CI: -0.62, 1.36 and SMD= 0.00; 95% CI: -0.98, 0.98 respectively)
Godges <i>et al.</i> , 1993 <sup>61</sup>	16 healthy, athletic, male college students, 21.00±1.00 years	n=9, static stretching, hip flexors, 3 sets, 2 minutes, 2 sessions per week, 3 weeks	n=7, continue their current activity levels	No significant difference between groups in gait economy (SMD= 0.83; 95% CI: -0.21, 1.87)

SMD: standard mean difference, CI: confidence interval, 10MWT: 10-meter walk test, 6MWT: 6-minute walk test, ROM: ROM, ?: information not provided.

**Table 2.** Quality assessment of the included studies.

Study	Eligibility criteria specified	Random allocation	Concealed allocation	Groups similar at baseline	Participant blinding	Therapist blinding	Assessor blinding	<15% dropouts	Intention-to-treat analysis	Between-group difference reported	Point estimate and variability reported	Total score	Patient category
Kerrigan <i>et al.</i> , 2003	y	y	n	y	n	n	y	y	n	y	y	6	Healthy older adults
Gajdosik <i>et al.</i> , 2005	y	y	y	y	n	n	n	y	y	y	y	7	
Christiansen, 2008	y	y	n	y	n	n	n	y	n	y	y	5	
Cristopoliski <i>et al.</i> , 2009	y	y	n	y	n	n	n	y	y	y	y	6	
Watt <i>et al.</i> , 2011	y	y	n	n	n	n	y	n	n	n	y	3	
Locks <i>et al.</i> , 2012	y	n	n	y	n	n	n	n	n	y	y	3	
Watt <i>et al.</i> , 2011	y	y	n	?	n	n	y	n	n	n	y	3	Frail older adults
Hotta <i>et al.</i> , 2009	y	y	n	y	n	n	n	y	n	y	y	5	Peripheral artery disease
Kim <i>et al.</i> , 2013	n	n	n	y	n	n	n	n	n	y	y	3	Stroke
Johanson <i>et al.</i> , 2006	y	y	n	n	n	n	n	y	y	y	y	5	Lower limb overuse injury
Johanson <i>et al.</i> , 2009	y	y	n	y	n	n	n	y	y	y	y	6	Limited ankle ROM
Godges <i>et al.</i> , 1993	y	y	n	?	n	n	n	y	y	y	y	5	Healthy adults

n: criterion not fulfilled; y: criterion fulfilled; ?: criterion not mentioned; total score: each item (except the first) contributes 1 point to the total score, yielding a PEDro scale score that can range from 0 to 10.

### Meta-analyses

Four meta-analysis were conducted for the following outcomes (Figure 2): gait speed, stride length, hip extension during gait and anterior pelvic tilt.

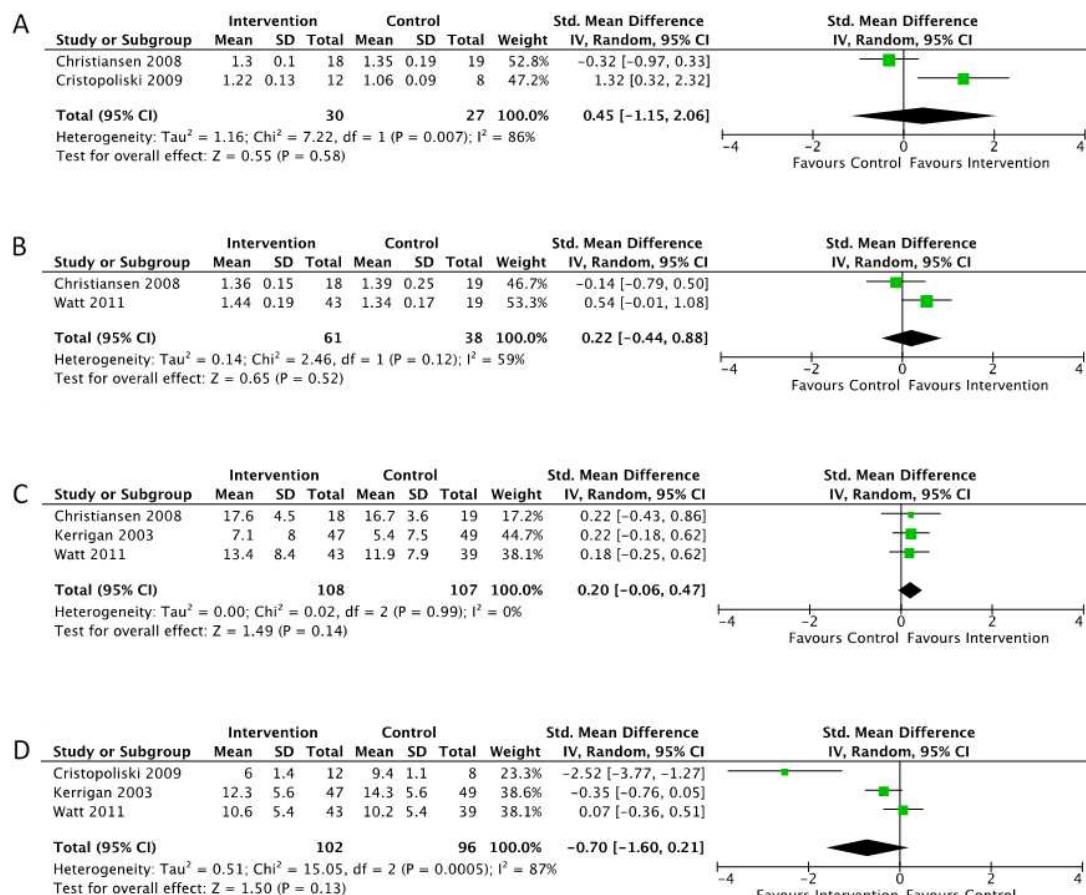
**Gait speed:** For gait speed (Figure 2A), two studies were included in the meta-analysis<sup>51,52</sup>. One study was excluded because intervention and control groups were not similar at baseline<sup>55</sup>. Statistical analysis showed no significant difference between groups (SMD= 0.45; 95% CI: -1.15, 2.06), with heterogeneous results ( $I^2=86%$ ,  $p=0.007$ ). Thus, the level of evidence was conflicting.

**Stride length:** For stride length (Figure 2B), two studies were included in the meta-analysis<sup>51,55</sup>. Statistical analysis showed no significant difference between groups (SMD= 0.22; 95% CI: -0.44, 0.88), with consistent results ( $I^2=59%$ ,  $p=0.12$ ). Only one study was of high quality<sup>51</sup>, thus a moderate level of

evidence supports the lack of beneficial effect of stretching to improve stride length in the elderly.

**Hip extension:** For hip extension during gait (kinematic data) (Figure 2C), three studies were included in the meta-analysis<sup>51,53,55</sup>. Statistical analysis showed no significant difference between groups (SMD= 0.20; 95% CI: -0.06, 0.47), with consistent results ( $I^2=0%$ ,  $p=0.99$ ). Two studies were of high quality<sup>51,53</sup>, thus a strong level of evidence supports the lack of beneficial effect of stretching to improve hip ROM during gait in the elderly.

**Anterior pelvic tilt:** For anterior pelvic tilt (Figure 2D), three studies were included in the meta-analysis<sup>52,53,55</sup>. Statistical analysis showed no significant difference between groups (SMD= -0.70; 95% CI: -1.60, 0.21), with heterogeneous results ( $I^2=87%$ ,  $p<0.01$ ). Thus, the level of evidence was conflicting.



**Figure 2.** Comparisons between intervention and control groups for gait speed (A), stride length (B), hip extension (C) and anterior pelvic tilt (D) in healthy older adults.



**Effects of interventions in other outcomes**

For the outcomes below, no meta-analysis could be performed because only one study was identified. Nevertheless, for each outcome, effect size estimates with a measure of statistical uncertainty (95% CI) were provided.

**Angular variables during gait initiation:** The study of Christiansen *et al.* (2008) showed no significant difference between stretching and control groups for ankle dorsiflexion during gait (SMD=0.29; 95% CI: -0.36, 0.94) with a moderate level of confidence (PEDro score: 5/10). The study of Kerrigan *et al.* (2003) showed no significant difference between groups for ankle plantar flexion (SMD=-0.05; 95% CI: -0.45, 0.35), with a moderate level of confidence (PEDro score: 6/10). The study of Cristopoliski *et al.* (2009) showed no significant difference between groups for lateral pelvic tilt (SMD= 0.93; 95% CI: -0.02, 1.88) and knee ROM (SMD= 0.23; 95% CI: -0.67, 1.12), with a moderate level of confidence (PEDro score: 6/10).

**Kinetic variables:** The study of Kerrigan *et al.* (2003) showed no significant difference between groups for hip torque (SMD= 0.35; 95% CI: -0.06, 0.75) and ankle plantar flexion power (SMD=0.00; 95% CI: -0.40, 0.40), with a moderate level of confidence (PEDro score: 6/10).

**Spatiotemporal variables:** The study of Cristopoliski *et al.* (2009) showed no significant difference between groups for cycle duration (SMD= -0.24; 95% CI: -1.14, 0.66), heel contact velocity (SMD= -0.46; 95% CI: -1.37, 0.45) and toe clearance (SMD= 0.91; 95% CI: -0.04, 1.86). However, the study showed significant decreases with large effect sizes in stance phase duration (SMD=-1.92; 95% CI: -3.04, -0.81), double support phase duration (SMD= -1.69; 95% CI: -2.76, -0.62) in favour of the stretching group as compared to the control group. Additionally, the authors found significant increases with large effect sizes of swing phase duration (SMD=1.92; 95% CI: 0.81, 3.04) and step length (SMD=1.37; 95% CI: 0.36, 2.38) in favour of the stretching group as compared to the control group. The study obtained a PEDro score of 6/10, thus, the level of evidence for these outcomes was moderate.

**Functional tests:** The study of Gajdosik *et al.* (2005) showed no significant difference between groups for the 10MWT (SMD= -0.76; 95% CI= -1.70, 0.18), with a moderate level of confidence (PEDro score: 7/10). The study of Locks *et al.* (2012) showed no significant improvement of the 6MWT in favour of the stretching group as compared to the control group (SMD= -0.04; 95% CI:-0.86, 0.79) with a limited level of confidence (low quality CCT with a PEDro score of 3/10).

**Frail elderly****Description of the study and quality assessment**

The study of Watt *et al.* 2011 examined the effects of stretching on frail elderly subjects<sup>56</sup>. Regarding the characteristics of the subjects, 74 subjects were included, and the mean age was 77.0±8.0 years. Regarding the characteristics of the training

programs, the stretching program lasted ten weeks, with a frequency of 14 sessions per week (two sessions per day). Participants performed two sets per session, holding the stretch for 60 seconds (static stretching), alternating the right and left limb (four minutes in total). The muscle group stretched was the hip flexors. The outcomes were cadence (steps/minute), walking speed (meters/second), stride length (meters) peak hip extension (degree) and peak anterior pelvic tilt (degree). Regarding the quality assessment, the study was identified as RCT and had an average PEDro score of 3 (low level of evidence).

**Effects of intervention**

The study of Watt *et al.* (2011) showed no significant difference between groups in angular variables, i.e. peak hip extension and anterior pelvic tilt (SMD= 0.22; 95% CI: -0.24, 0.68 and SMD= -0.05; 95% CI: -0.51, 0.41 respectively). There was also no significant difference for cadence (SMD= 0.13; 95% CI: -0.33, 0.59). However, the study showed significant improvements in favour of the stretching group with small effect sizes in some performance-related variables, i.e. walking speed and stride length (both SMD= 0.49; 95% CI: 0.03, 0.96).

**Elderly with symptomatic peripheral artery disease****Description of the study and quality assessment**

The study of Hotta *et al.* (2019) examined the effects of stretching in elderly with symptomatic peripheral artery disease<sup>57</sup>. Regarding the characteristics of the subjects, 13 subjects were included and the mean age was not mentioned. Regarding the characteristics of the training programs, the stretching program lasted four weeks, with a frequency of five sessions per week. Participants performed one set daily, holding the stretch for 30 minutes (static stretching with splints). The muscle group stretched was ankle plantar flexors. The gait outcome was 6MWT. Regarding the quality assessment, the study was identified as RCT and had an average PEDro score of 5 (moderate level of evidence).

**Effects of intervention**

The study of Hotta *et al.* (2019) showed significant improvements in favour of the stretching group for both total walking distance and continuous walking distance with large effect sizes (SMD= 1.56; 95% CI: 0.66, 2.45 and SMD= 3.05; 95% CI: 1.86, 4.23 respectively).

**Stroke****Description of the study and quality assessment**

The study of Kim *et al.* (2013) examined the effects of stretching on stroke patients<sup>58</sup>. Only a static muscle stretching training group and control group were included in the analysis. Regarding the characteristics of the subjects, 24 subjects were included, and the mean age was 53.3±3.1 years. Regarding the characteristics of the training programs, the stretching program lasted six weeks, with a frequency of four sessions per week. Participants performed one set per session, holding the stretch for 20 minutes (static stretching). The muscle group stretched was ankle plantar flexors. The outcome was the sway of the centre of pressure during the stance phase.

Regarding the quality assessment, the study was identified as CCT and had an average PEDro score of 3 (low level of evidence).

#### Effects of intervention

The study of Kim *et al.* (2013) showed no significant difference between groups in the sway of the centre of pressure (SMD=0.75; 95% CI: -0.09, 1.58).

#### Young adults with limited ankle ROM and a history of lower limb overuse injury

##### Description of the study and quality assessment

The study of Johanson *et al.* (2006) examined the effects of stretching on healthy adults with limited passive ankle-dorsiflexion ROM (less than eight degrees) and a history of lower limb overuse injury<sup>59</sup>. Regarding the characteristics of the subjects, 19 subjects were included and the mean age was 30.3±9.8 years. Regarding the characteristics of the training programs, the stretching program lasted three weeks, with a frequency of two sessions per day. Participants performed five sets per session, holding the stretch for 30 seconds (static stretching). The muscle group stretched was ankle plantar flexors. The outcomes were ankle dorsiflexion and time-to-heel-off during the stance phase of gait. Regarding the quality assessment, the study was identified as RCT and had an average PEDro score of 5 (moderate level of evidence).

#### Effects of intervention

The study of Johanson *et al.* (2006) showed no significant difference between groups in ankle dorsiflexion during gait in both the right and left ankle (SMD= 0.50; 95% CI: -0.42, 1.43 and SMD= 0.41; 95% CI: -0.52, 1.33 respectively). There was also no significant difference between groups for time-to-heel-off during the stance phase of gait in both the right and left ankle (SMD= -0.50; 95% CI: -1.43, 0.43 and SMD= -0.48; 95% CI: -1.41, 0.45 respectively).

#### Young adults with limited ankle ROM

##### Description of the study and quality assessment

The study of Johanson *et al.* (2009) examined the effects of stretching on young adults with limited passive ankle-dorsiflexion ROM (less than 5 degrees)<sup>60</sup>. It is worth noticed that these participants were not the same than in the study of Johanson *et al.* (2006). In contrast, the characteristics of the training programs were the same as in Johanson *et al.* (2006). In the current study, 16 subjects were included, and the mean age was 27.4±8.2 years. The muscle group stretched was the ankle plantar flexors. The outcomes were maximum ankle dorsiflexion, maximum knee extension and EMG amplitude of the gastrocnemius during the stance phase of gait. Regarding the quality assessment, the study was identified as RCT and had an average PEDro score of 6 (moderate level of evidence).

#### Effects of intervention

The study of Johanson *et al.* (2009) showed no significant difference between groups in angular variables during gait, i.e. maximum ankle dorsiflexion and maximum knee extension (SMD= 0.53; 95% CI: -0.48, 1.53 and SMD= -0.07; 95% CI: -1.05, 0.91 respectively). There was also no significant

difference between groups for EMG variables, i.e. medial and lateral gastrocnemius activity (SMD= 0.37; 95% CI: -0.62, 1.36 and SMD= 0.00; 95% CI: -0.98, 0.98 respectively).

#### Healthy young adults

##### Description of the study and quality assessment

The study of Godges *et al.* (1993) examined the effects of stretching on healthy young adults<sup>61</sup>. Only a static hip extension stretching group and control group were included in the analysis. Regarding the characteristics of the subjects, 16 subjects were included, and the mean age was 21.0±1.0 years. Regarding the characteristics of the training programs, the stretching program lasted three weeks, with a frequency of two sessions per week. Participants performed three sets per session, holding the stretch for two minutes (static stretching). The muscle group stretched was the hip flexors. The outcome was walking economy (ml/kg/min). Regarding the quality assessment, the study was identified as RCT and had an average PEDro score of 5 (moderate level of evidence).

#### Effects of intervention

The study of Godges *et al.* (1993) showed no significant difference between groups in gait economy in terms of oxygen consumption (SMD= 0.83; 95% CI: -0.21, 1.87).

#### Discussion

The aim of this systematic review was to determine the effects of a stretching program on human gait by means of a systematic literature review and meta-analysis. Twelve studies were identified in six different patient categories. Statistical analyses showed no strong level of evidence supporting the beneficial effect of a stretching program to improve any gait outcome. The major issue in conducting meta-analyses and establishing strong level of evidences was the great heterogeneity in gait variables. The results obtained in the different patient categories are discussed in detail below.

#### Healthy older adults

The healthy older adult population was the most studied. Two muscle groups were systematically stretched in the six identified studies: hip flexors<sup>23,51-53,55</sup> and ankle plantar flexors<sup>23,51-53</sup>. Hip flexor stiffness, associated with reduced hip extension during gait has been demonstrated in the elderly and may alter gait<sup>62,63</sup>. In the same way, decreased calf muscle length associated with restricted dorsiflexion ROM is well documented in older adults<sup>35,64</sup>. A decreased ankle dorsiflexion ROM has been correlated with poorer balance test scores in the elderly<sup>65</sup> and may contribute to an increased risk of falls<sup>66</sup>. All the studies included in the present analysis showed that specific stretching programs were efficient to improve passive ROM of the targeted joints, but results are more heterogeneous regarding gait performance and dynamic ROM. This led to inconsistency in the results or the impossibility to conclude with a strong level of evidence that a stretching program improves gait in healthy older adults. Moreover, when improvement in ROM or gait performance occurred, it was not associated with a significant increase in dynamic hip extension or ankle dorsiflexion. Only trends toward increased dynamic ROM after stretching interventions were

observed<sup>51,53,55</sup>. This observation was consistent in young adults.

When data were meta-analyzed, we ensured that the groups and the training characteristics were similar to limit the risk of bias. This explains that a limited number of studies was included in the meta-analysis. It is worth noticed that the stretching technic was the same (i.e. static stretching), but that details of interventions varied across these studies. For example, both studies selected for the meta-analysis of gait speed included hip flexors and plantar flexors stretching, but, one study included hip extensor stretching<sup>52</sup> whereas the other did not<sup>51</sup>. This difference may partially explain the heterogenous results in the meta-analysis ( $I^2=86\%$ ,  $p=0.007$ ). In the same way, the heterogenous results observed in the meta-analysis of anterior pelvic tilt ( $I^2=87\%$ ,  $p<0.01$ ) may be explained by the stretching of additional muscle groups (hip extensors and plantar flexors) in the study of Cristopoliski *et al.* (2009) compared to the other studies (in which only hip flexors were stretched)<sup>53,55</sup>. Nevertheless, heterogeneity in the results was not systematically observed between studies that used slight different protocols, as showed by the consistent results in the meta-analyses of stride length ( $I^2=59\%$ ,  $p=0.12$ ) and hip extension ( $I^2=0\%$ ,  $p=0.99$ ). Thus, we assume that we have limited the risk of bias in the meta-analyses.

### Stroke patients

In stroke patients, ankle plantar flexor stretching has been successfully used to improve ankle stiffness<sup>67-70</sup>. Decreased plantar flexors stiffness may have a beneficial effect on postural control during gait because *triceps surae* is known to play an important role during gait<sup>71-73</sup> and an increase in muscle stiffness might alter synergistic muscle activities during human gait. However, only one non-randomized study<sup>58</sup> was identified and included in the current systematic review. Other studies that used stretching in multicomponent programs<sup>74-76</sup> or in control groups<sup>77,78</sup> were identified but excluded because of the addition of resistance training or the lack of a control group. Nevertheless, it should be noted that some studies showed improvements between pre- and post-stretching conditions. Forrester *et al.* (2014) showed that both robotic ankle mobilizations and manual ankle stretching improved gait velocity in stroke patients at hospital discharge compared to baseline. Similarly, Park *et al.* (2018) showed that both static ankle stretching and ankle mobilizations improved gait speed after four weeks of treatment compared to baseline. Other authors showed that one week of immobilization in dorsiflexed position (casting) followed by one week of plantar flexor stretching and gait training improved gait performances in 10MWT and 6MWT<sup>74</sup>. Hence, these encouraging results suggest that further randomized controlled trials of good quality are needed to explore the ability of ankle stretching to improve gait parameters in stroke or in other neurological diseases exposing patients to joint stiffness, e.g. Parkinson's disease<sup>79</sup>.

### Young adults

In healthy adults, the interest of practicing stretching to improve gait seems limited as they are assumed to have sufficient mobility for walking. Moreover, the included study involved athletic males<sup>61</sup>, a population that is known to be more

flexible than inactive persons<sup>80</sup>. Stretching should be more indicated when ROM is limited<sup>24</sup>. However, even in young adults with limited ankle ROM, stretching did not improve dynamic dorsiflexion during gait<sup>59,60</sup>. Stretching programs in apparently healthy adults should be more indicated after a prolonged period of reduced functional demand (e.g. immobilization, sedentarity), when ROM is insufficient to practice a specific activity or when high levels of flexibility are required for sport performance (e.g. gymnastics or dance) and in sports that involve stretch-shortening cycles (e.g. basketball, volleyball)<sup>24</sup>.

### Limitations of the study

Some patient categories were not included in the present review, although muscle stretching is commonly indicated in their clinical care to reduce spasticity<sup>81</sup>. This is for example the case for children with cerebral palsy<sup>82</sup>. In fact, we were able to identify studies in the literature focusing on the effects of stretching on gait in this population during the first phase of the present review, but the protocol of these studies combined stretching with another form of training (e.g. 83,84) or there was no control group (e.g. 85). These studies therefore did not fit with the inclusion criteria of the present systematic review and were consequently excluded. Now, it should be stressed that the effectiveness of static stretching to improve motor function in children with cerebral palsy is still controversial<sup>86</sup>, although some authors showed that functional stretching exercises may be effective to improve gait<sup>85</sup>. Further randomized controlled trials are needed to explore the impact of stretching on gait in this population.

To reduce the risk of bias, data were meta-analyzed when at least 2 studies with similar populations and training characteristics were found. Considering these constraints and the great heterogeneity in the gait outcomes, we only performed meta-analyses in healthy older adults for 4 outcomes. The Cochrane Qualitative and Implementation Methods Group recommends the application of Grades of Recommendation, Assessment, Development, and Evaluation (GRADE) in the Evidence from Qualitative Reviews to assess the level of confidence in systematic review with meta-analyses<sup>87</sup>. However, the GRADE necessitates assessing the risk of publication bias with a funnel plot, determining its asymmetry, which can be performed with at least 10 studies<sup>88</sup>. In the present study, the meta-analyses included less than 10 randomized controlled trials, so, we chose to implement other guidelines described by a Cochrane collaboration group to assess the level of evidence<sup>50</sup>. Because this method includes fewer criteria, our confidence in the results must be taken with caution.

### Conclusion

Twelve studies were identified, involving a total of 442 subjects. Despite some improvements in isolated studies, statistical analyses showed no strong level of evidence supporting the beneficial effect of using stretching alone to improve gait outcomes in rehabilitation programs. The major obstacle in conducting meta-analyses and establishing strong levels of evidence were the great heterogeneity in gait variables and the low quality of the included studies. Because the effects of stretching are not clear, further randomized controlled trials of good

quality are needed to understand the impact of stretching on human gait. Currently, stretching is more recommended to maintain and improve ROM rather than improve gait parameters and should be integrated in multicomponent programs.

### Data availability

#### Underlying data

All data underlying the results are available as part of the article and no additional source data are required.

### Reporting guidelines

Harvard Dataverse: PRISMA checklist and PRISMA flow diagram for 'Effects of stretching exercises on human gait: a systematic review and meta-analysis', <https://doi.org/10.7910/DVN/N8ZXNB><sup>89</sup>.

Data are available under the terms of the [Creative Commons Zero "No rights reserved" data waiver](#) (CC0 1.0 Public domain dedication).

### References

- Beauchet O, Allali G, Sekhon H, et al.: **Guidelines for Assessment of Gait and Reference Values for Spatiotemporal Gait Parameters in Older Adults: The Biomathics and Canadian Gait Consortiums Initiative.** *Front Hum Neurosci.* 2017; **11**: 353.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
- Dicharry J: **Kinematics and kinetics of gait: from lab to clinic.** *Clin Sports Med.* 2010; **29**(3): 347–364.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Ounpuu S: **The biomechanics of walking and running.** *Clin Sports Med.* 1994; **13**(4): 843–863.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Tinetti ME: **Performance-oriented assessment of mobility problems in elderly patients.** *J Am Geriatr Soc.* 1986; **34**(2): 119–126.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Podsiadlo D, Richardson S: **The Timed "Up & Go": A Test of Basic Functional Mobility for Frail Elderly Persons.** *J Am Geriatr Soc.* 1991; **39**(2): 142–148.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Yiou E, Artico R, Teysseire CA, et al.: **Anticipatory Postural Control of Stability during Gait Initiation Over Obstacles of Different Height and Distance Made Under Reaction-Time and Self-Initiated Instructions.** *Front Hum Neurosci.* 2016; **10**: 449.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
- Yiou E, Fourcade P, Artico R, et al.: **Influence of temporal pressure constraint on the biomechanical organization of gait initiation made with or without an obstacle to clear.** *Exp Brain Res.* 2016; **234**(6): 1363–1375.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Yiou E, Teysseire C, Artico R, et al.: **Comparison of base of support size during gait initiation using force-plate and motion-capture system: A Bland and Altman analysis.** *J Biomech.* 2016; **49**(16): 4168–4172.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Artico R, Fourcade P, Teysseire CA, et al.: **Influence of swing foot strike pattern on balance control mechanisms during gait initiation over an obstacle to clear.** In: *Progress in Motor Control*, Amsterdam, Netherlands. 7–10 Jul 2019; 2019.  
[Reference Source](#)
- LeBrasseur NK: **Gait as an Integrative Measure and Predictor of Health Across Species.** *J Gerontol A Biol Sci Med Sci.* 2019; **74**(9): 1411–1412.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Blumen HM, Cavallari P, Mourey F, et al.: **Editorial: Adaptive Gait and Postural Control: from Physiological to Pathological Mechanisms, Towards Prevention and Rehabilitation.** *Front Aging Neurosci.* 2020; **12**: 45.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
- Yiou E, Hamaoui A, Allali G: **Editorial: The Contribution of Postural Adjustments to Body Balance and Motor Performance.** *Front Hum Neurosci.* 2018; **12**: 487.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
- White DK, Neogi T, Nevitt MC, et al.: **Trajectories of Gait Speed Predict Mortality in Well-Functioning Older Adults: The Health, Aging and Body Composition Study.** *J Gerontol A Biol Sci Med Sci.* 2013; **68**(4): 456–464.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
- Rodríguez-Moliner A, Herrero-Larrea A, Miñarro A, et al.: **The spatial parameters of gait and their association with falls, functional decline and death in older adults: a prospective study.** *Sci Rep.* 2019; **9**(1): 8813.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
- Allum JH, Bloem BR, Carpenter MG, et al.: **Proprioceptive control of posture: a review of new concepts.** *Gait Posture.* 1998; **8**(3): 214–242.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Horak FB: **Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls?** *Age Ageing.* 2006; **35** Suppl 2: ii7–ii11.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Mirelman A, Shema S, Maidan I, et al.: **Gait.** *Handb Clin Neurol.* 2018; **159**: 119–134.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Arcolin I, Pisano F, Delconte C, et al.: **Intensive cycle ergometer training improves gait speed and endurance in patients with Parkinson's disease: A comparison with treadmill training.** *Restor Neural Neurosci.* 2015; **34**(1): 125–138.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Lustosa LP, Silva JP, Coelho FM, et al.: **Impact of resistance exercise program on functional capacity and muscular strength of knee extensor in pre-frail community-dwelling older women: a randomized crossover trial.** *Rev Bras Fisioter.* 2011; **15**(4): 318–324.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Halvarsson A, Franzén E, Ståhle A: **Balance training with multi-task exercises improves fall-related self-efficacy, gait, balance performance and physical function in older adults with osteoporosis: a randomized controlled trial.** *Clin Rehabil.* 2015; **29**(4): 365–375.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Fischer M, Vialleron T, Laffaye G, et al.: **Long-Term Effects of Whole-Body Vibration on Human Gait: A Systematic Review and Meta-Analysis.** *Front Neurol.* 2019; **10**: 627.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
- Freiberger E, Häberle L, Spirduso WW, et al.: **Long-term effects of three multicomponent exercise interventions on physical performance and fall-related psychological outcomes in community-dwelling older adults: a randomized controlled trial.** *J Am Geriatr Soc.* 2012; **60**(3): 437–446.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Gajdosik RL, Vander Linden DW, McNair PJ, et al.: **Effects of an eight-week stretching program on the passive-elastic properties and function of the calf muscles of older women.** *Clin Biomech (Bristol, Avon).* 2005; **20**(9): 973–983.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Portero P, McNair P: **Les étirements musculotendineux : des données scientifiques à une pratique raisonnée.** *Kinésithérapie, la Revue.* 2015; **15**(164–165): 32–40.  
[Publisher Full Text](#)
- Delafontaine A, Gagey O, Colnaghi S, et al.: **Rigid Ankle Foot Orthosis Deteriorates Mediolateral Balance Control and Vertical Braking during Gait Initiation.** *Front Hum Neurosci.* 2017; **11**: 214.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
- Delafontaine A, Fourcade P, Honeine JL, et al.: **Postural adaptations to unilateral knee joint hypomobility induced by orthosis wear during gait initiation.** *Sci Rep.* 2018; **8**(1): 830.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
- Delafontaine A, Honeine JL, Do MC, et al.: **Comparative gait initiation kinematics between simulated unilateral and bilateral ankle hypomobility: Does bilateral constraint improve speed performance?** *Neurosci Lett.* 2015; **603**: 55–59.  
[PubMed Abstract](#) | [Publisher Full Text](#)
- Alamini-Rodrigues C, Hamaoui A: **Effect of three different lumbar splints on**

- posturokinetic capacity when performing the sit-to-stand task. *Ann Phys Rehabil Med.* 2017; **60**(6): 406–409.  
[PubMed Abstract](#) | [Publisher Full Text](#)
29. Hamaoui A, Alami-Rodrigues C: **Influence of Cervical Spine Mobility on the Focal and Postural Components of the Sit-to-Stand Task.** *Front Hum Neurosci.* 2017; **11**: 129.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
30. Grimston SK, Nigg BM, Hanley DA, et al.: **Differences in ankle joint complex range of motion as a function of age.** *Foot Ankle.* 1993; **14**(4): 215–222.  
[PubMed Abstract](#) | [Publisher Full Text](#)
31. James B, Parker AW: **Active and passive mobility of lower limb joints in elderly men and women.** *Am J Phys Med Rehabil.* 1989; **68**(4): 162–167.  
[PubMed Abstract](#) | [Publisher Full Text](#)
32. McKay MJ, Baldwin JN, Ferreira P, et al.: **Normative reference values for strength and flexibility of 1,000 children and adults.** *Neurology.* 2017; **88**(1): 36–43.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
33. Roach KE, Miles TP: **Normal hip and knee active range of motion: the relationship to age.** *Phys Ther.* 1991; **71**(9): 656–665.  
[PubMed Abstract](#) | [Publisher Full Text](#)
34. Soucie JM, Wang C, Forsyth A, et al.: **Range of motion measurements: reference values and a database for comparison studies.** *Haemophilia.* 2011; **17**(3): 500–507.  
[PubMed Abstract](#) | [Publisher Full Text](#)
35. Vandervoort AA, Chesworth BM, Cunningham DA, et al.: **Age and sex effects on mobility of the human ankle.** *J Gerontol.* 1992; **47**(1): M17–21.  
[PubMed Abstract](#) | [Publisher Full Text](#)
36. Lee PG, Jackson EA, Richardson CR: **Exercise Prescriptions in Older Adults.** *Am Fam Physician.* 2017; **95**(7): 425–432.  
[PubMed Abstract](#)
37. Mora JC, Valencia WM: **Exercise and Older Adults.** *Clin Geriatr Med.* 2018; **34**(1): 145–162.  
[PubMed Abstract](#) | [Publisher Full Text](#)
38. Beyaert C, Vasa R, Frykberg GE: **Gait post-stroke: Pathophysiology and rehabilitation strategies.** *Neurophysiol Clin.* 2015; **45**(4–5): 335–355.  
[PubMed Abstract](#) | [Publisher Full Text](#)
39. Abbruzzese G, Marchese R, Avanzino L, et al.: **Rehabilitation for Parkinson's disease: Current outlook and future challenges.** *Parkinsonism Relat Disord.* 2016; **22 Suppl 1**: S60–S64.  
[PubMed Abstract](#) | [Publisher Full Text](#)
40. Bisht B, Darling WG, White EC, et al.: **Effects of a multimodal intervention on gait and balance of subjects with progressive multiple sclerosis: a prospective longitudinal pilot study.** *Degener Neurol Neuromuscul Dis.* 2017; **7**: 79–93.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
41. Thong-On S, Bovonsunthonchai S, Vachalathiti R, et al.: **Effects of Strengthening and Stretching Exercises on the Temporospacial Gait Parameters in Patients With Plantar Fasciitis: A Randomized Controlled Trial.** *Ann Rehabil Med.* 2019; **43**(6): 662–676.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
42. van Lith BJH, den Boer J, van de Warrenburg BPC, et al.: **Functional effects of botulinum toxin type A in the hip adductors and subsequent stretching in patients with hereditary spastic paraplegia.** *J Rehabil Med.* 2019; **51**(6): 434–441.  
[PubMed Abstract](#) | [Publisher Full Text](#)
43. Behm DG, Chaouachi A: **A review of the acute effects of static and dynamic stretching on performance.** *Eur J Appl Physiol.* 2011; **111**(11): 2633–2651.  
[PubMed Abstract](#) | [Publisher Full Text](#)
44. Kay AD, Blazevich AJ: **Effect of acute static stretch on maximal muscle performance: a systematic review.** *Med Sci Sports Exerc.* 2012; **44**(1): 154–164.  
[PubMed Abstract](#) | [Publisher Full Text](#)
45. Moher D, Liberati A, Tetzlaff J, et al.: **Preferred Reporting Items for Systematic Reviews and Meta-Analyses: The PRISMA Statement.** *PLoS Med.* 2009; **6**(7): e1000097.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
46. de Morton NA: **The PEDro scale is a valid measure of the methodological quality of clinical trials: a demographic study.** *Aust J Physiother.* 2009; **55**(2): 129–133.  
[PubMed Abstract](#) | [Publisher Full Text](#)
47. Lindberg J, Carlsson J: **The effects of whole-body vibration training on gait and walking ability - A systematic review comparing two quality indexes.** *Physiother Theory Pract.* 2012; **28**(7): 485–98.  
[PubMed Abstract](#) | [Publisher Full Text](#)
48. Higgins JP, Green S: **Cochrane Handbook for Systematic Reviews of Interventions.** Accessed March 26, 2019.  
[Reference Source](#)
49. Cohen J: **Statistical Power Analysis for the Behavioral Sciences.** 2nd ed. L. Erlbaum Associates; 1988.  
[Publisher Full Text](#)
50. van Tulder M, Furlan A, Bombardier C, et al.: **Updated Method Guidelines for Systematic Reviews in the Cochrane Collaboration Back Review Group.** *Spine (Phila Pa 1976).* 2003; **28**(12): 1290–1299.  
[PubMed Abstract](#) | [Publisher Full Text](#)
51. Christiansen CL: **The effects of hip and ankle stretching on gait function of older people.** *Arch Phys Med Rehabil.* 2008; **89**(8): 1421–1428.  
[PubMed Abstract](#) | [Publisher Full Text](#)
52. Cristopoliski F, Barela JA, Leite N, et al.: **Stretching Exercise Program Improves Gait in the Elderly.** *Gerontology.* 2009; **55**(6): 614–620.  
[PubMed Abstract](#) | [Publisher Full Text](#)
53. Kerrigan DC, Xenopoulos-Oddsson A, Sullivan MJ, et al.: **Effect of a hip flexor-stretching program on gait in the elderly.** *Arch Phys Med Rehabil.* 2003; **84**(1): 1–6.  
[PubMed Abstract](#) | [Publisher Full Text](#)
54. Locks RR, Costa TC, Koppe S, et al.: **Effects of strength and flexibility training on functional performance of healthy older people.** *Rev Bras Fisioter.* 2012; **16**(3): 184–190.  
[PubMed Abstract](#) | [Publisher Full Text](#)
55. Watt JR, Jackson K, Franz JR, et al.: **Effect of a Supervised Hip Flexor Stretching Program on Gait in Elderly Individuals.** *PM R.* 2011; **3**(4): 324–329.  
[PubMed Abstract](#) | [Publisher Full Text](#)
56. Watt JR, Jackson K, Franz JR, et al.: **Effect of a Supervised Hip Flexor Stretching Program on Gait in Frail Elderly Patients.** *PM R.* 2011; **3**(4): 330–335.  
[PubMed Abstract](#) | [Publisher Full Text](#)
57. Hotta K, Batchelor WB, Graven J, et al.: **Daily Passive Muscle Stretching Improves Flow-Mediated Dilation of Popliteal Artery and 6-minute Walk Test in Elderly Patients with Stable Symptomatic Peripheral Artery Disease.** *Cardiovasc Revasc Med.* 2019; **20**(8): 642–648.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
58. Kim TH, Yoon JS, Lee JH: **The Effect of Ankle Joint Muscle Strengthening Training and Static Muscle Stretching Training on Stroke Patients' C.O.P Sway Amplitude.** *J Phys Ther Sci.* 2013; **25**(12): 1613–1616.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
59. Johanson MA, Wooden M, Catlin PA, et al.: **Effects of gastrocnemius stretching on ankle dorsiflexion and time-to heel-off during the stance phase of gait.** *Phys Ther Sport.* 2006; **7**(2): 93–100.  
[Publisher Full Text](#)
60. Johanson MA, Cuda BJ, Koontz JE, et al.: **Effect of stretching on ankle and knee angles and gastrocnemius activity during the stance phase of gait.** *J Sport Rehabil.* 2009; **18**(4): 521–534.  
[PubMed Abstract](#) | [Publisher Full Text](#)
61. Godges JJ, MacRae PG, Engelke KA: **Effects of Exercise on Hip Range of Motion, Trunk Muscle Performance, and Gait Economy.** *Phys Ther.* 1993; **73**(7): 468–477.  
[PubMed Abstract](#) | [Publisher Full Text](#)
62. Kerrigan DC, Lee LW, Collins JJ, et al.: **Reduced hip extension during walking: healthy elderly and fallers versus young adults.** *Arch Phys Med Rehabil.* 2001; **82**(1): 26–30.  
[PubMed Abstract](#) | [Publisher Full Text](#)
63. Kerrigan DC, Todd MK, Della Croce U, et al.: **Biomechanical gait alterations independent of speed in the healthy elderly: evidence for specific limiting impairments.** *Arch Phys Med Rehabil.* 1998; **79**(3): 317–322.  
[PubMed Abstract](#) | [Publisher Full Text](#)
64. Gajdosik RL, Linden VWD, Vander Linden DW: **Influence of Age on Length and Passive Elastic Stiffness Characteristics of the Calf Muscle-Tendon Unit of Women.** *Phys Ther.* 1999; **79**(9): 827–838.  
[PubMed Abstract](#) | [Publisher Full Text](#)
65. Mecagni C, Smith JP, Roberts KE, et al.: **Balance and ankle range of motion in community-dwelling women aged 64 to 87 years: a correlational study.** *Phys Ther.* 2000; **80**(10): 1004–1011.  
[PubMed Abstract](#) | [Publisher Full Text](#)
66. Gehlsen GM, Whaley MH: **Falls in the elderly: Part II Balance, strength, and flexibility.** *Arch Phys Med Rehabil.* 1990; **71**(10): 739–741.  
[PubMed Abstract](#)
67. Gao F, Ren Y, Roth EJ, et al.: **Effects of repeated ankle stretching on calf muscle-tendon and ankle biomechanical properties in stroke survivors.** *Clin Biomech (Bristol, Avon).* 2011; **26**(5): 516–522.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
68. Yeh CY, Tsai KH, Chen JJ: **Effects of prolonged muscle stretching with constant torque or constant angle on hypertonic calf muscles.** *Arch Phys Med Rehabil.* 2005; **86**(2): 235–241.  
[PubMed Abstract](#) | [Publisher Full Text](#)
69. Yeh CY, Chen JJ, Tsai KH: **Quantitative analysis of ankle hypertonia after prolonged stretch in subjects with stroke.** *J Neurosci Methods.* 2004; **137**(2): 305–314.  
[PubMed Abstract](#) | [Publisher Full Text](#)
70. Bressel E, McNair PJ: **The effect of prolonged static and cyclic stretching on ankle joint stiffness, torque relaxation, and gait in people with stroke.** *Phys Ther.* 2002; **82**(9): 880–887.  
[PubMed Abstract](#) | [Publisher Full Text](#)
71. Honeine JL, Schieppati M, Gagey O, et al.: **By counteracting gravity, triceps surae sets both kinematics and kinetics of gait.** *Physiol Rep.* 2014; **2**(2): e00229.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
72. Honeine JL, Schieppati M, Gagey O, et al.: **The Functional Role of the Triceps Surae Muscle during Human Locomotion.** *PLoS One.* 2013; **8**(1): e52943.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
73. Neptune RR, Kautz SA, Zajac FE: **Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation**

- during walking. *J Biomech.* 2001; **34**(11): 1387–1398.  
[PubMed Abstract](#) | [Publisher Full Text](#)
74. Carda S, Invernizzi M, Baricich A, *et al.*: **Casting, taping or stretching after botulinum toxin type A for spastic equinus foot: a single-blind randomized trial on adult stroke patients.** *Clin Rehabil.* 2011; **25**(12): 1119–1127.  
[PubMed Abstract](#) | [Publisher Full Text](#)
75. Ghasemi E, Khademi-Kalantari K, Khalkhali-Zavieh M, *et al.*: **The effect of functional stretching exercises on functional outcomes in spastic stroke patients: A randomized controlled clinical trial.** *J Bodyw Mov Ther.* 2018; **22**(4): 1004–1012.  
[PubMed Abstract](#) | [Publisher Full Text](#)
76. Moore SA, Jakovljevic DG, Ford GA, *et al.*: **Exercise Induces Peripheral Muscle But Not Cardiac Adaptations After Stroke: A Randomized Controlled Pilot Trial.** *Arch Phys Med Rehabil.* 2016; **97**(4): 596–603.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
77. Forrester LW, Roy A, Krywonis A, *et al.*: **Modular Ankle Robotics Training in Early Sub-Acute Stroke: A Randomized Controlled Pilot Study.** *Neurorehabil Neural Repair.* 2014; **28**(7): 678–687.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
78. Park D, Lee JH, Kang TW, *et al.*: **Four-week training involving ankle mobilization with movement versus static muscle stretching in patients with chronic stroke: a randomized controlled trial.** *Top Stroke Rehabil.* 2019; **26**(2): 81–6.  
[PubMed Abstract](#) | [Publisher Full Text](#)
79. Raza C, Anjum R, Shakeel NUA: **Parkinson's disease: Mechanisms, translational models and management strategies.** *Life Sci.* 2019; **226**: 77–90.  
[PubMed Abstract](#) | [Publisher Full Text](#)
80. Haff G, Triplett NT, National Strength & Conditioning Association (U.S.), eds. **Essentials of Strength Training and Conditioning.** Fourth edition. Human Kinetics; 2016.  
[Reference Source](#)
81. Smania N, Picelli A, Munari D, *et al.*: **Rehabilitation procedures in the management of spasticity.** *Eur J Phys Rehabil Med.* 2010; **46**(3): 423–438.  
[PubMed Abstract](#)
82. Pin T, Dyke P, Chan M: **The effectiveness of passive stretching in children with cerebral palsy.** *Dev Med Child Neurol.* 2006; **48**(10): 855–862.  
[PubMed Abstract](#) | [Publisher Full Text](#)
83. Fosdahl MA, Jahnsen R, Kvalheim K, *et al.*: **Effect of a Combined Stretching and Strength Training Program on Gait Function in Children with Cerebral Palsy, GMFCS Level I & II: A Randomized Controlled Trial.** *Medicina (Kaunas).* 2019; **55**(6): 250.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
84. Kalkman BM, Holmes G, Bar-On L, *et al.*: **Resistance Training Combined With Stretching Increases Tendon Stiffness and Is More Effective Than Stretching Alone in Children With Cerebral Palsy: A Randomized Controlled Trial.** *Front Pediatr.* 2019; **7**: 333.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
85. Elshafey MA, Abd-Elaziem A, Gouda RE: **Functional stretching exercise submitted for spastic diplegic children: a randomized control study.** *Rehabil Res Pract.* 2014; **2014**: 814279.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
86. Kalkman BM, Bar-On L, O'Brien TD, *et al.*: **Stretching Interventions in Children With Cerebral Palsy: Why Are They Ineffective in Improving Muscle Function and How Can We Better Their Outcome?** *Front Physiol.* 2020; **11**: 131.  
[PubMed Abstract](#) | [Publisher Full Text](#) | [Free Full Text](#)
87. Noyes J, Booth A, Flemming K, *et al.*: **Cochrane Qualitative and Implementation Methods Group guidance series-paper 3: methods for assessing methodological limitations, data extraction and synthesis, and confidence in synthesized qualitative findings.** *J Clin Epidemiol.* 2018; **97**: 49–58.  
[PubMed Abstract](#) | [Publisher Full Text](#)
88. Higgins J, Green S: **Cochrane Handbook for Systematic Reviews of Interventions Version 5.1.0 [Updated March 2011].** The Cochrane Collaboration, 2011.  
[Reference Source](#)
89. Delafontaine A: **Effects of stretching exercises on human gait: A systematic review and meta-analysis.** Harvard Dataverse, V1. 2020.  
<http://www.doi.org/10.7910/DVN/N8ZXNB>

### **3.2.2. Effets immédiats des étirements du triceps sural sur la mobilité de la cheville et l'organisation posturale lors de l'initiation de la marche chez des patients atteints de la maladie de Parkinson : étude randomisée contrôlée**

Cette étude a donné lieu à la publication ci-dessous :

**Vialleron T**, Delafontaine A, Millerioux I, Memari S, Fourcade P, Yiou E (2021). Acute effects of short-term stretching of the *triceps surae* on ankle mobility and gait initiation in patients with Parkinson's disease. *Clinical Biomechanics* 89: 105449. <https://doi.org/10.1016/j.clinbiomech.2021.105449>.



## Original Articles

Acute effects of short-term stretching of the *triceps surae* on ankle mobility and gait initiation in patients with Parkinson's diseaseThomas Vialleron<sup>a,b</sup>, Arnaud Delafontaine<sup>a,b</sup>, Isis Millerioux<sup>a,b</sup>, Sahel Memari<sup>a,b</sup>, Paul Fourcade<sup>a,b</sup>, Eric Yiou<sup>a,b,\*</sup><sup>a</sup> CIAMS laboratory, Université Paris-Saclay, 91405 Orsay Cedex, France<sup>b</sup> CIAMS laboratory, Université d'Orléans, 45067 Orléans, France

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## ABSTRACT

**Background:** Ankle mobility is known to be of uttermost importance to generate propulsive forces and control balance during gait initiation. Impaired mobility of the postural chain occurs with normal ageing and is exacerbated in patients with Parkinson's disease. This study questions whether short-term stretching session applied to the *triceps surae* improves ankle mobility and, consequently, dynamical postural control in patients with Parkinson's disease performing gait initiation.

**Method:** Nineteen patients with Parkinson's disease participated in this study and were randomly assigned to an "intervention group" or a "sham group". In the intervention group, patients were exposed to a 4 × 60 seconds *triceps surae* stretching. In the sham group, they were exposed to forearm stretching. Additionally, ten age-matched healthy elderly, who were not exposed to any stretching-treatment, were assigned to a "control group". Participants performed series of gait initiation on a force-plate before and after their treatment.

**Findings:** Ankle mobility was improved in the intervention group after *triceps surae* stretching. The forward velocity of the center-of-mass at heel-off and motor performance related-parameters (progression velocity, center-of-mass velocity at foot-contact and swing phase duration) were also improved in the intervention group, with large effect sizes ( $d \geq 0.8$ ). None of the stability parameters were modified by the treatments.

**Interpretation:** Short-term *triceps surae* stretching is an efficient method to increase ankle mobility and improve the capacity to generate forward propulsive forces in patients with Parkinson's disease. These findings are congruent with the "posturo-kinetics capacity" theory according to which dynamical postural control depends on postural chain mobility.

## 1. Introduction

Gait initiation (GI), the transient period between quiet standing posture and steady-state walking (Brenière and Do, 1986), is a locomotor task that is widely used in the literature to investigate postural control in patients with Parkinson's disease (e.g. Bonora et al., 2017; Carpinella et al., 2007; Chastan et al., 2009; Halliday et al., 1998; Mancini et al., 2009; Okada et al., 2011a, 2011b). GI can be divided in two successive phases: a postural phase preceding the swing heel-off, where dynamic phenomena corresponding to the so-called "anticipatory postural adjustments" (APA) are generated, followed by the execution phase (Brenière and Do, 1986). Along the anteroposterior axis, APA include a backward center-of-pressure (COP) displacement which promotes the initial forward propulsive forces required to reach

the intended step length and walking speed (Brenière et al., 1987; Lepers and Brenière, 1995). This postural dynamics is due to bilateral inhibition of ankle plantar-flexors activity, followed by strong bilateral activation of ankle dorsi-flexors (Crenna and Frigo, 1991). Along the mediolateral axis, APA include COP displacement towards the swing leg which promotes center-of-mass (COM) displacement in the opposite direction, i.e., towards the stance leg. This postural dynamic acts to attenuate the COM fall towards the swing leg during the execution phase and therefore contributes to stability maintenance (Yiou et al., 2016a). This anticipatory mediolateral COP shift has been classically attributed to the loading of the swing leg associated with the activation of swing leg hip abductors (Carlsöö, 1966; Winter, 1995). Recent study further reported that, the stance knee and hip are slightly flexed during APA (Honeine et al., 2016) which acts to unload the ipsilateral leg and therefore

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complement the action of swing hip abductors. EMG analysis revealed that the flexion of the stance knee is favored by bilateral *soleus* silencing and a greater ipsilateral *tibialis anterior* activity with respect to contralateral activity, while stance hip flexion is associated with activation of the stance *rectus femoris*. Finally, during GI execution phase, the activation of the *TS* of the stance leg has been shown to brake the vertical COM fall under gravity, which facilitates stability maintenance after foot-contact (Honeine et al., 2013; Honeine et al., 2014). This brief literature review on the postural organization of GI thus stresses the key role of the ankle musculature for balance control and motor performance (for a complete review, see (Yiou et al., 2017)).

This postural organization of GI is known to be altered in patients with Parkinson's disease (PD). There is however no clear consensus on the effect of PD on GI at its early stage. For example, Mancini et al. (2009) found that the amplitude of APA was altered in the mediolateral direction, but not in the anteroposterior direction. Carpinella et al. (2007) did not investigate the mediolateral APA but found that the anteroposterior component of APA was altered in early PD patients. Differences in the initial posture and/or inclusion criteria may have explained these discrepancies (Mancini et al., 2009). Later in the disease, it seems clear from the literature that the amplitude of both mediolateral and anteroposterior APA becomes bradykinetic (Halliday et al., 1998). In addition, advanced PD patients exhibit altered capacity to actively brake the COM fall (Chastan et al., 2009). These motor impairments contribute to gait disorders in this population and are exacerbated in patients with the freezing of gait or without medication (Mancini et al., 2020).

These deficits in postural control in PD patients can be envisaged from the point of view of the biomechanical concept of "Posturo-kinetics Capacity" (PKC) (e.g. (Bazalgette et al., 1987); for a complete review on the PKC concept, see (Bouisset and Do, 2008). According to this concept, any factor that alters the dynamic mobility of the postural chain, would negatively impact APA, and would therefore alter motor performance and/or stability. This dynamic mobility would indeed be essential to generate the external forces required to accelerate the COM during APA and brake the COM fall during the GI execution phase. This concept was substantiated by recent experimental studies which investigated the relationship between lower limb mobility, APA, motor performance and stability in young healthy adults who are performing GI under experimental joint hypomobility induced by orthosis wear (Delafontaine et al., 2017, 2015; Delafontaine et al., 2018).

Chronic joint hypomobility or rigidity is a recurring symptom of PD (Halliday et al., 1998). This rigidity is characterized by an excessive and continuous muscle tone which produces stiffness and resistance to movement in joints (Raza and Anjum, 2019). Given the above considerations on the PKC concept and the brief literature review provided above, such increased joint rigidity - especially of the ankle - would hinder APA development and/or COM braking, and may contribute to stability impairments and slowness of movement performance, which are typical motor symptoms of PD patients. In line, recent study showed that the wear of a rigid ankle foot orthosis in healthy young adults induced an alteration of both APA and COM braking during GI (Delafontaine et al., 2017). According to the PKC concept, it can thus be expected that improving ankle mobility with specific treatments would facilitate APA development and/or COM braking, with a consequent positive effect on motor performance and/or stability. Ankle muscle stretching, an intervention known to increase ankle mobility in healthy elderly (Gajdosik et al., 2005) and that can easily be applied by patients themselves, may represent such a treatment. The present study was thus designed to test the hypothesis that a short-term stretching session applied to the *Triceps surae* (*TS*) improves ankle mobility and, consequently, the postural organization of GI in PD patients.

## 2. Method

### 2.1. Participants

Nineteen PD patients participated in this study (see Table 1 for details on demographic data of participants). They were randomly assigned to an "intervention group" or a "sham group". Ten age-matched healthy older adults additionally participated in this study and were assigned to the "control group". They were community dwelling seniors, free of any known neuromuscular disorders and able to walk without assistance. They were included in the experiment to provide normative values of dependent variables. PD patients were examined in their OFF-medication state (i.e., after 12-h withdrawal from antiparkinson's medications, as defined by Defer et al., 1999) by an experienced neurophysiotherapist. The examination was based on the Movement Disorder Society Clinical Diagnostic Criteria for Parkinson's disease (Postuma et al., 2015). PD patients had no history of falling, were physically independent, were able to walk without assistance, and did not experience freezing. All participants gave written informed consent after being instructed about the nature and purpose of the experiment, which was approved by the local ethics committee of the University of Paris-Saclay. This study complied with the standards established by the Declaration of Helsinki and was assigned the following trial registration number: 2018-A00162-53.

### 2.2. Ankle range of motion (ROM) assessment

An experienced physical therapist, who was blinded to the group assignment, measured each participant's maximum passive ankle dorsiflexion in supine position, with the foot in subtalar neutral position and the knee straight. This measure was made with a manual goniometer (Merz Pharma, France) as it is known to be reliable for assessing ankle dorsiflexion ROM (Konor et al., 2012; Martin and McPoil, 2005). The axis of the goniometer was aligned with the lateral malleolus of the ankle. The proximal arm was aligned with the head of the fibula. The distal arm was aligned with the lateral border of the fifth metatarsal (Webber and Porter, 2010). A right angle between these two arms corresponded to 0° dorsiflexion. The final position of dorsiflexion corresponded to the point at which subjects felt pain or discomfort, or the experimenter perceived resistance to stretch. Two measurements were taken for each leg and then averaged to determine the passive dorsiflexion of each ankle (Webber and Porter, 2010). The ankle dorsiflexion of both legs was assessed for each participant in the intervention and sham group before and immediately after a stretching treatment (cf. below). Statistical analysis revealed that the ankle dorsiflexion of the most affected side (cf. Table 1) was not significantly different to that of the least affected side. For clarity's sake in the report of data, values of both legs were therefore pooled.

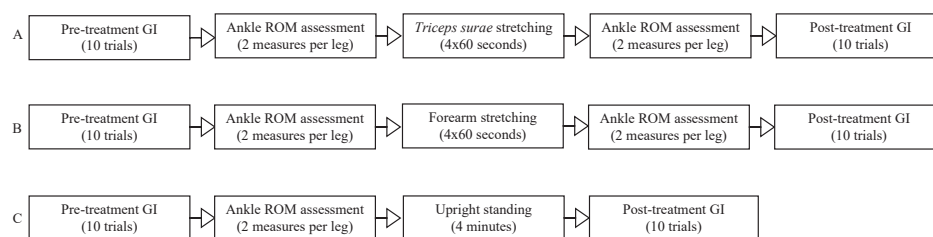
### 2.3. Treatment

In the intervention group, patients stood upright with both hands against a wall, and with both feet on a wedge (20° tilt) in order to stretch their *TS*. They were instructed to slightly lean forward while keeping their heels in contact with the wedge and their body aligned until they felt the stretch of the calf without pain. They were then asked to maintain their posture. In the sham group, patients were instructed to position their arm outstretched, palm facing the front, extending their wrist and fingers by using their other hand (both arms were outstretched). In both groups, series of four stretches lasting 60 s each were performed (Shrier and Gossal, 2000). In the control group, no stretching was carried out. Participants simply stood upright for the same duration as for the stretching treatments (Fig. 1).

**Table 1**

Demographic data of patients with Parkinson's disease. M, male; F, female; R, right; L, left; MDS-UPDRS, Movement Disorder Society–Unified Parkinson's Disease Rating Scale; SD, standard deviation.

	Subjects	Genre	Age (years)	Height (meters)	Body mass (kilograms)	Disease Durations (years)	Most disabled side	Hoehn and Yahr state	MDS-UPDRS (part 3)
Intervention group	1	M	77	1.58	62	2	R	2	18
	2	F	70	1.74	70	8	L	3	28
	3	M	76	1.70	62	5	R	2	14
	4	F	63	1.69	76	7	R	2	10
	5	M	71	1.70	76	13	L	2	16
	6	M	67	1.70	70	6	L	2	14
	7	M	66	1.58	45	5	L	2	36
	8	F	77	1.73	82	5	R	2	14
	9	F	69	1.76	87	2	R	2	35
	10	M	57	1.60	66	9	L	2	26
Mean (SD)	–	–	69.30 (6.45)	1.68 (0.07)	69.60 (11.89)	6.20 (3.29)	–	2.10 (0.32)	21.10 (9.41)
Sham group	1	M	68	1.60	52	4	R	2	17
	2	M	57	1.55	48	2	L	2	35
	3	M	63	1.57	40	4	L	2	21
	4	F	71	1.80	90	6	R	3	50
	5	M	69	1.70	82	2	L	2	32
	6	M	74	1.59	51	12	R	3	20
	7	F	70	1.75	74	8	L	2	20
	8	F	56	1.78	90	3	L	2	43
	9	F	74	1.65	65	14	L	2	36
	10	M	69	1.78	74	–	–	–	–
Mean (SD)	–	–	66.89 (6.75)	1.67 (0.10)	65.78 (19.02)	6.11 (4.37)	–	2.22 (0.44)	30.44 (11.63)
Control group	1	F	70	1.71	52	–	–	–	–
	2	F	78	1.55	65	–	–	–	–
	3	F	69	1.64	55	–	–	–	–
	4	M	79	1.78	67	–	–	–	–
	5	F	67	1.60	86	–	–	–	–
	6	F	65	1.62	71	–	–	–	–
	7	F	74	1.60	54	–	–	–	–
	8	F	76	1.62	56	–	–	–	–
	9	F	69	1.75	68	–	–	–	–
	10	M	69	1.78	74	–	–	–	–
Mean (SD)	–	–	71.10 (4.56)	1.67 (8.30)	64.90 (10.86)	–	–	–	–



**Fig. 1.** Study protocol. A, intervention group; B, sham group; C, control group; GI, gait initiation; ROM, range of motion.

#### 2.4. Experimental task

Participants initially stood barefoot in a natural upright posture on a force-plate (0.9 × 1.80 m, AMTI, Watertown, USA) embedded at the beginning of a 6-m walkway track. They performed two series of ten GI: one series just before, and a second series immediately after the treatment (pre-treatment and post-treatment condition, respectively; Fig. 1). They were instructed to initiate gait at a spontaneous velocity when they felt ready, after receiving an “all set” signal, and continue walking until the end of the track. The rest time was ~10 s between trials.

#### 2.5. Raw data

Force-plate data were low-pass filtered using a second order Butterworth filter with a 10 Hz cut-off frequency (Honeine et al., 2014). The mediolateral and anteroposterior coordinates of the COP were computed from force-plate data. The instantaneous COM acceleration was

determined from the ground reaction forces according to the Newton's second law. COM velocity and displacement were computed by successive numerical integrations of COM acceleration. Data were collected at a rate of 500 Hz. Data acquisition was controlled by a custom-made program written in Matlab™ (Version 5.3 (R11), The MathWorks Inc., USA).

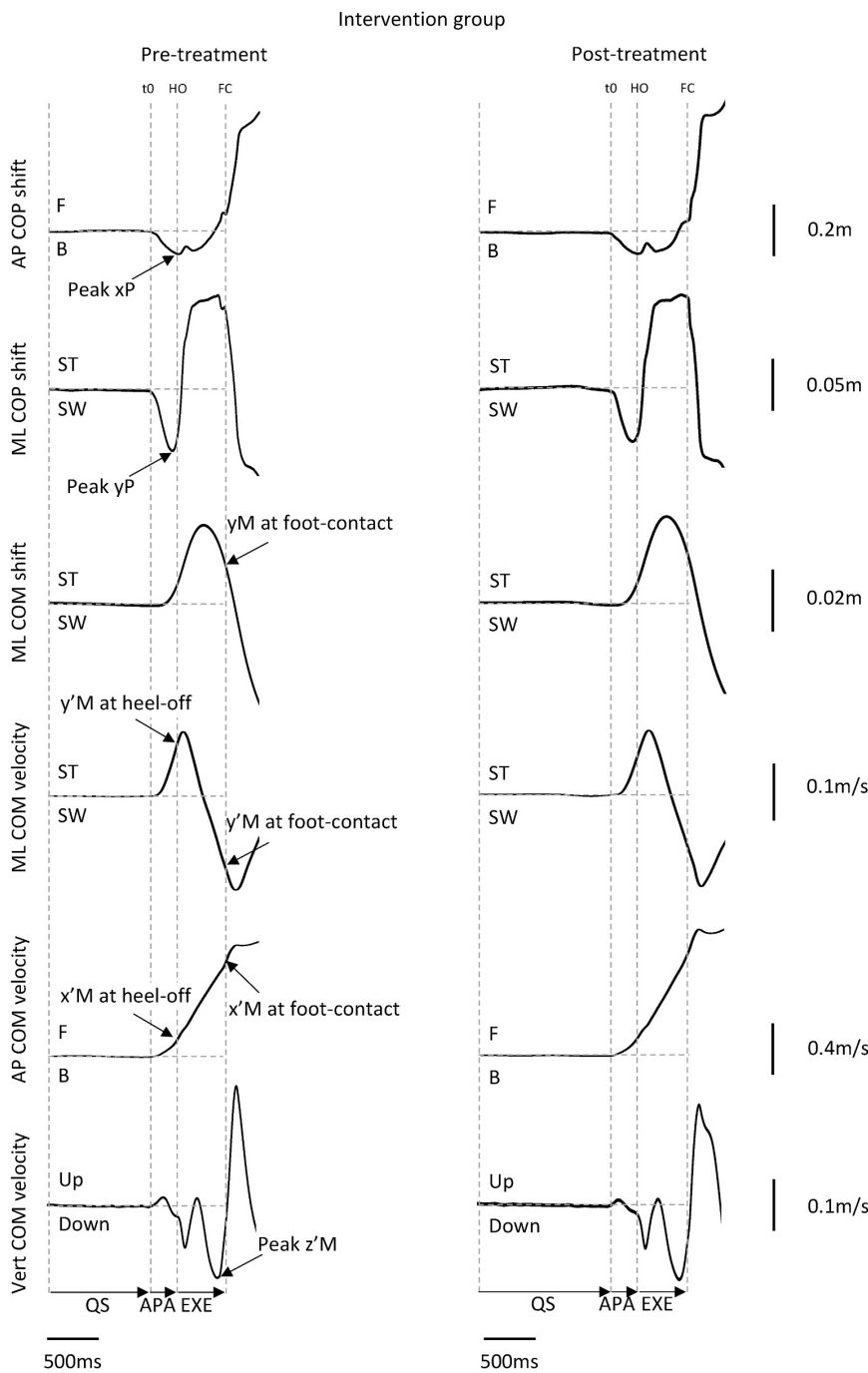
#### 2.6. GI variables

##### 2.6.1. Time-events

The following time-events were determined from the biomechanical traces (Fig. 2): GI onset ( $t_0$ ), swing heel-off and swing foot-contact (cf. (Caderby et al., 2013), for the determination of these instants; cf. Fig. 2).

##### 2.6.2. Initial posture, APA and execution phase

The COM position in the initial posture was estimated by averaging the COP position values during the 250 ms period preceding the “all set”



**Fig. 2.** Example of typical biomechanical traces of gait initiation obtained in one representative subject of the intervention group, and representation of the main experimental variables. AP, ML, Vert: anteroposterior, mediolateral and vertical; xP, yP: AP and ML center of pressure (COP) shift; yM: ML center of mass (COM) shift; x'M, y'M, z'M: AP, ML and vertical COM velocity; F, B: forward and backward; ST, SW: stance and swing leg side; Vertical lines: t0, onset variation of biomechanical traces; HO, heel-off; FC, foot-contact. Horizontal arrows: QS, quiet standing; APA, anticipatory postural adjustments; EXE, execution phase.

signal. GI was divided into APA (from t0 to heel-off) and execution phase (from heel-off to foot-contact, (Brenière et al., 1987)). APA amplitude was characterized with the peak COP displacement obtained during the APA time-window and the COM velocity at swing heel-off.

### 2.6.3. Motor performance

Motor performance was quantified with the anteroposterior COM

velocity at foot-contact, the mean progression velocity ( $v$ ), with  $v = [\text{step length}] / [\text{duration of the execution phase}]$  (Brenière and Do, 1986), step length and execution phase duration.

### 2.6.4. Mediolateral stability

An adaptation of the “margin of stability” (MOS) introduced by Hoff et al. (Hof et al., 2005) was used to quantify mediolateral dynamic

stability at foot-contact. The MOS corresponded to the difference between the mediolateral boundary of the base of support ( $BOS_{y_{max}}$ ) and the mediolateral position of the “extrapolated COM” at swing foot-contact ( $YCOM_{FC}$ ). Thus:

$$MOS = BOS_{y_{max}} - YCOM_{FC}$$

$BOS_{y_{max}}$  was estimated with the mediolateral COP position at the time of rear foot-off (Caderby et al., 2017; You et al., 2016b). The mediolateral position of the extrapolated COM at foot-contact ( $yCOM_{FC}$ ) was calculated as follows:

$$yCOM_{FC} = yM_{FC} + \frac{y'M_{FC}}{\omega_0}$$

where  $yM_{FC}$  and  $y'M_{FC}$  are respectively the mediolateral COM position and velocity at foot-contact, and  $\omega_0$  is the eigenfrequency of the body, modeled as an inverted pendulum and calculated as follows:

$$\omega_0 = \sqrt{\frac{g}{l}}$$

where  $g = 9.81 \text{ m/s}^2$  is the gravitational acceleration and  $l$  is the length of the inverted pendulum, which in this study corresponded to 57.5% of body height (Winter et al., 1990). Mediolateral dynamic stability at foot-contact is preserved on the condition that  $yCOM_{FC}$  is within  $BOS_{y_{max}}$ , which corresponds to a positive MOS. A negative MOS indicates a mediolateral instability and implies that a corrective action is required to maintain balance.

### 2.6.5. Vertical COM braking

The “braking index” introduced by Do and colleagues (e.g. (Chastan et al., 2009)) provides evidence that the central nervous system prepares for foot-contact by decreasing the COM vertical velocity to achieve a soft landing (Honeine et al., 2013; Honeine et al., 2014). It was calculated as:

$$\text{Braking index} = \frac{(V_{min} - V_{FC})}{V_{min}}$$

where  $V_{min}$  is the minimal vertical COM velocity and  $V_{FC}$  is the vertical COM velocity at foot-contact.

## 2.7. Statistics

Mean values and standard deviations were calculated for each dependent variable. For ankle dorsiflexion, a repeated-measures ANOVA model with one within-subject factor and one between-subject factor was used. The within-subject factor was the time of the treatment (or “time” factor; 2 levels: pre-treatment, post-treatment) and the between-subject factor was the group (2 levels: sham, intervention). Additionally, a simple *t*-test was used to compare ankle dorsiflexion between PD patients and healthy participants in the pre-treatment condition. For GI parameters, a repeated-measures ANOVA model was also used. The within-subject factor was the time (2 levels: pre-treatment, post-treatment) and the between-subject factor was the group (3 levels: control, sham, intervention). The Tukey post hoc test was used when required. The threshold of significance was set at  $P < 0.05$ . The Cohen’s *d* was used to assess the effect sizes when repeated-measures ANOVA showed a significant time X group interaction. The effect sizes were classified as trivial ( $<0.2$ ), small (0.2–0.49), medium (0.5–0.79), and large ( $\geq 0.8$ ).

## 3. Results

### 3.1. Biomechanical traces

The time course of the biomechanical traces was globally similar in the different groups and time conditions. The traces obtained in a representative patient of the intervention group are reported in Fig. 2.

### 3.2. Ankle ROM

There was a significant effect of time ( $F(1,17) = 8.54, P < 0.05$ ) and a significant time X group interaction ( $F(1,17) = 8.54, P < 0.05$ ) on the ankle dorsiflexion ROM. Post hoc analysis revealed that it was significantly larger in the post-treatment than in the pre-treatment condition for the intervention group only ( $P < 0.01, d = 0.97$ ). In the pre-treatment condition, PD patients had a significantly lower ankle ROM compared to healthy participants ( $P < 0.05, d = 0.74$ ).

### 3.3. Initial posture and APA

There was no significant effect of time and group, nor any significant time X group interaction on the following variables: initial COM position, APA duration, anteroposterior peak COP displacement and mediolateral COM velocity at heel-off. In contrast, there was a significant effect of time ( $F(2,26) = 15.28, P < 0.001$ ) and a significant time X group interaction ( $F(2,26) = 5.64, P < 0.01$ ) on the anteroposterior COM velocity at heel-off. Post hoc analysis revealed that it was significantly larger in the post-treatment than in the pre-treatment condition for the intervention group only ( $P < 0.001, d = 1.17, \text{Fig. 3}$ ). Post hoc analysis further showed that it was significantly larger for the intervention group in the post-treatment condition than for the control group in both the pre- and post-treatment conditions ( $P < 0.05, d = 1.27$  and  $d = 1.33$ , respectively). Finally, there was a significant effect of group on the mediolateral peak of COP displacement ( $F(2,26) = 4.62, P < 0.05$ ), with a larger value obtained for the control than for both PD groups.

### 3.4. Motor performance

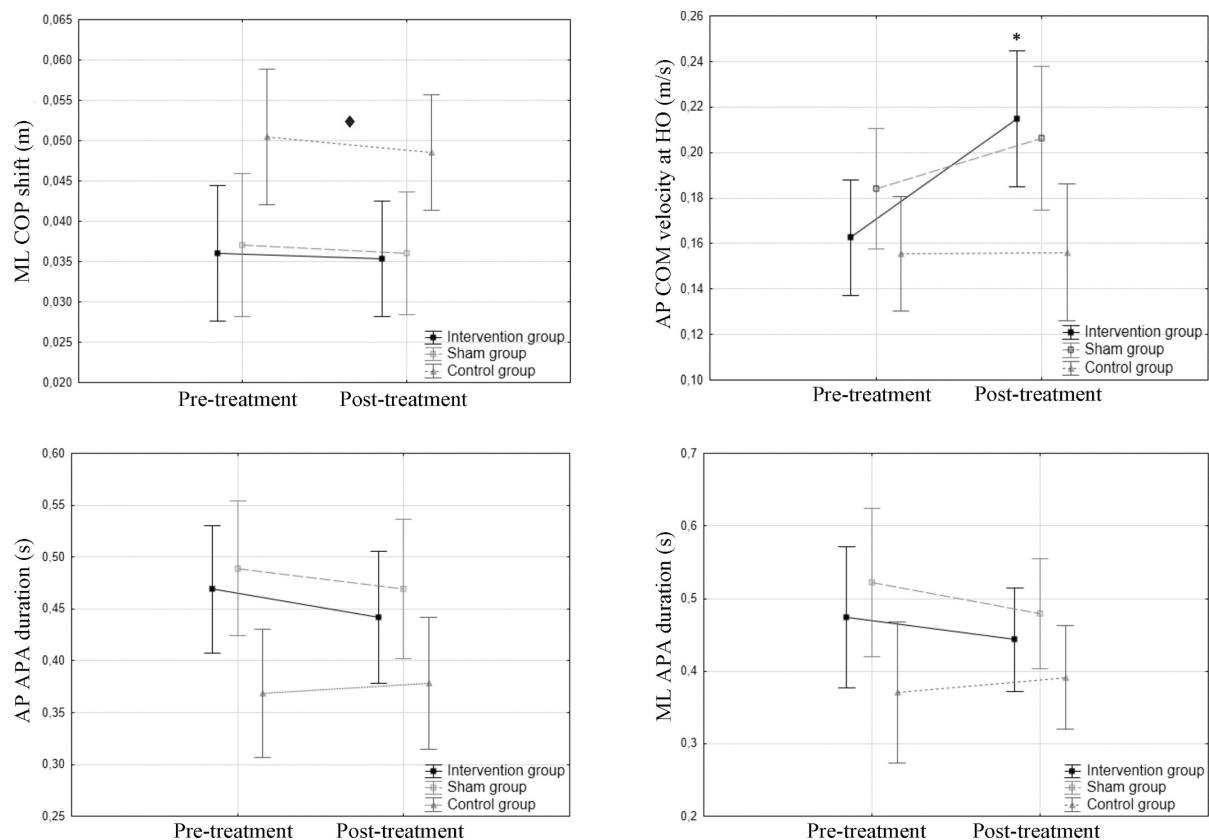
There was no significant effect of group on the anteroposterior COM velocity at foot-contact, but a significant effect of time ( $F(1,26) = 19.58, P < 0.001$ ) and a significant time X group interaction ( $F(2,26) = 5.40, P < 0.05$ ). This interaction could be ascribed to a significantly larger COM velocity in the post-treatment than in the pre-treatment condition for the intervention ( $P < 0.01$ ) and the sham group ( $P < 0.05, d = 0.78$  and  $d = 0.59$ , respectively; Fig. 4). Similarly, there was no significant effect of group on the mean progression velocity, but a significant effect of time ( $F(1,26) = 14.05, P < 0.001$ ) and a significant time X group interaction ( $F(2,26) = 4.03, P < 0.05$ ). This variable was significantly larger in the post-treatment than in the pre-treatment condition for the intervention group only ( $P < 0.01, d = 1.07$ ). There was also a significant effect of time ( $F(1,26) = 8.00, P < 0.05$ ) and a time X group interaction ( $F(2,26) = 5.01, P < 0.05$ ) on the execution phase duration. The interaction could be ascribed to a significantly lower execution phase duration in the post-treatment than in the pre-treatment condition for the intervention group only ( $P < 0.01, d = 1.27$ ). Finally, there was no significant effect of time, group nor any time X group interaction on step length.

### 3.5. Mediolateral stability and vertical COM braking

There was a significant effect of group on the mediolateral COM position at foot-contact ( $F(2,26) = 30.10, P < 0.001$ ) and the MOS ( $F(2,26) = 13.78, P < 0.001$ ), but no significant effect of time nor time X group interaction on these variables. The COM position at foot-contact was larger for the control than for both PD groups, indicating that the COM fall towards the swing leg during the execution phase was smaller for the control group. This resulted in a larger MOS for the control group than for both PD groups (Fig. 5). Finally, there was no significant effect of time, group nor any significant time X group interaction on step width, mediolateral COM velocity at foot-contact and vertical COM braking.

## 4. Discussion

Impaired joint mobility is known to occur with normal ageing



**Fig. 3.** Effects of group and time on selected parameters of anticipatory postural adjustments (APA). COP, COM: center of pressure, center of mass; AP, ML: anteroposterior, mediolateral; HO: heel-off; vertical bars denote 0.95 confidence interval; ◆ indicates a significant effect of the group as revealed with the ANOVA; \* indicates a significant difference between the pre- and post-treatment condition within one group as revealed with the Tukey *post-hoc* test.

(Vandervoort et al., 1992) and is exacerbated in PD patients (Halliday et al., 1998). Previous reports showed that short-term *TS* stretching is an efficient training method to increase the ankle mobility in both young adults and healthy elderly (Nakamura et al., 2017). Results of the present study show that it is also efficient to improve ankle mobility in PD patients. The increase reached 44%.

Results further showed that several components of dynamical postural control were improved following *TS* stretching. Specifically, the anteroposterior COM velocity at heel-off was increased in the intervention group only, and this increase reached 32%. Previous reports showed that larger anteroposterior COM velocity at heel-off can be reached at the expense of a lengthened APA duration (Brenière et al., 1987). However, in the present experiment, APA duration remained unchanged between the pre- and post-treatment conditions. PD patients in the intervention group were therefore able to develop a larger quantity of initial forward propulsive forces during an equivalent time, which indicates that their capacity to generate forward APA was enhanced following the stretching. The improvement of motor performance reported in this group (in terms of increased mean progression velocity, increased COM velocity at foot-contact and reduced execution phase duration) can be seen as a consequence of this enhanced capacity to generate APA (Brenière et al., 1987).

Now, results further showed that, in contrast to all the other GI variables, the anteroposterior COM velocity at foot-contact was also increased in the sham group following the treatment. This finding suggests that part of the increase in this variable observed in the intervention group could probably be assigned to a placebo effect. Such

confounding effect was expected since it is known that PD patients are particularly influenced by the belief of improvement (Lidstone et al., 2010). However, this placebo effect only concerned the COM velocity at the time of foot-contact, and this increase was smaller in the sham than in the intervention group (12% vs. 8%), and so the effect size (0.78 vs. 0.59). The improvement in the postural organization of GI following *TS* stretching can therefore be considered as consistent.

Interestingly, PD patients who were exposed to the *TS* stretching reached an even larger anteroposterior COM velocity at heel-off than participants within the control group (healthy older adults). This finding can be ascribed to the relatively low-level of motor impairment of the PD patients included in the present study. All were able to walk without assistance, did not experience freezing, and had relatively low motor disorders according to the MDS-UPDRS and Hoehn & Yahr scale (cf. Table 1). Their locomotor function was therefore not much different from the healthy age-matched participants, as could also be reflected in the equivalent anteroposterior APA spatiotemporal parameters in the pre-treatment condition. As a consequence, increasing the initial COM velocity in these patients resulted in a significant difference with the control group in the post-treatment condition.

In fact, the effects of PD on the postural organization of GI only manifested as changes in the mediolateral stability-related parameters. Specifically, the amplitude of the mediolateral APA was lower in PD patients than in healthy participants. Previous study in healthy participants showed that the amplitude of these APA was predictive of the MOS value at foot-contact (You et al., 2016a). In the present study, the lower MOS value in the PD patients could therefore be ascribed to a lower

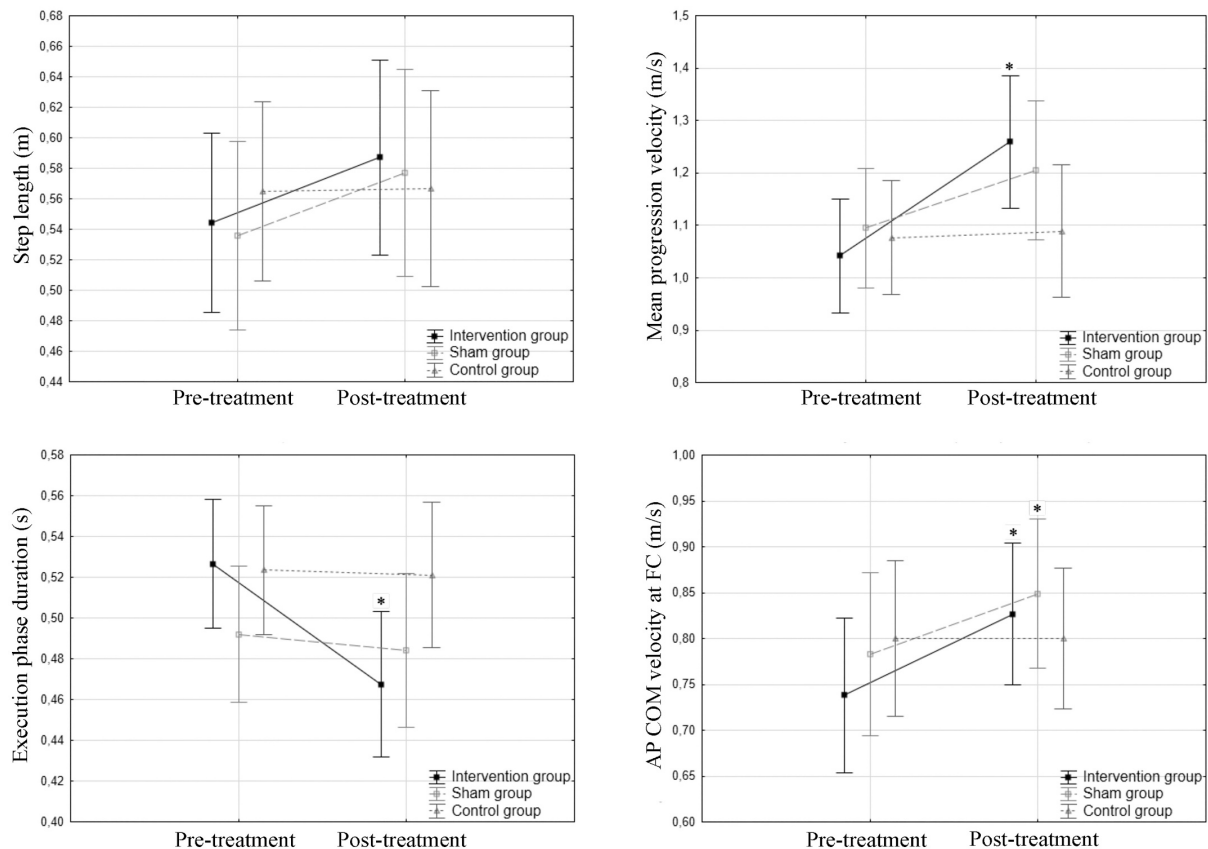


Fig. 4. Effects of group and time on selected parameters of motor performance. COM: center of mass; AP: anteroposterior; FC: foot contact; vertical bars denote 0.95 confidence interval; \* indicates a significant difference between the pre- and post-treatment condition within one group as revealed with the Tukey *post-hoc* test.

capacity of PD patients to generate mediolateral APA.

Our findings regarding the effect of PD on APA are in accordance with Mancini et al. (2009) who reported that, in early and untreated PD patients, the mediolateral component of the COP displacement during APA was altered, but not its anteroposterior component. Hypometric mediolateral APA may lead to an excessive weight shift towards the swing leg during the first step, as observed in PD by some authors (Okada et al., 2011a, 2011b), which increases the risk of imbalance in the frontal plane (Yiou et al., 2016a). Mancini et al. (2009) suggested that the pathology of PD may have a specific effect on loading/unloading of the legs early in the disease, consistent with abnormal mediolateral, but not anteroposterior, sway during quiet stance in PD patients (Mitchell et al., 1995). However, this assumption contrasts with the study of Carpinella et al. (2007) which reported a reduced backward COP displacement in early PD patients (phases 1 and 2 of Hoehn and Yahr scale). Differences in the initial posture and/or inclusion criteria may have explained this discrepancy (Mancini et al., 2009). For example, inclusion criteria for PD subjects were different, since all subjects in the study of Mancini et al. (2009) were untreated, whereas several subjects in the Carpinella's study were tested on dopaminergic medication. Despite this contradiction in early stage, it seems admitted that later in the disease, the magnitude of both mediolateral and anteroposterior APA becomes bradykinetic in PD (Halliday et al., 1998).

Similar to the anteroposterior APA, results of the present study showed that the capacity to actively brake the COM fall during GI was unaltered in PD patients. This finding contrasts with the study of Chastan et al. (2009) which reported that this capacity was altered in

advanced PD patients. Once again, a difference in the PD clinical stage may probably explain this discrepancy. Globally, these findings suggest that, in PD patients, the progressive decline in the control of mediolateral stability, vertical COM braking and forward propulsive forces may not follow the same evolution, maybe due to distinct systems within the basal ganglia circuitry. Short-term TS stretching did not improve the control of mediolateral stability, nor the capacity to brake the vertical COM fall. It is not excluded that more advanced PD patients than in our study, with more severe muscle rigidity, may benefit from TS stretching to improve these two parameters.

The present study has several limitations: 1) the PD patients involved in the experiment had relatively low-level of motor impairment. It is not known whether short-term stretching may also be beneficial to PD patients with more severe symptoms; 2) the sample sizes were small. We were however confident on the reliability of our results because of the large size effects found for most variables and the biomechanical coherence of data; 3) only short-term stretching effects were considered. To what extent PD patient might benefit from long-term stretching is unknown; 4) the carry-over effects were not evaluated.

In conclusion, the present results showed that short-term TS stretching substantially improved ankle mobility in PD patients. The capacity to generate forward propulsive forces during APA was also increased, with consequent improvement in motor performance. Now, it is obvious that PD motor impairment goes far beyond a simple alteration in ankle mobility. However, our results suggest that TS stretching might beneficially be integrated in multicomponent exercise programs, e.g. including strength, balance and gait, mental training (Shen et al., 2016),

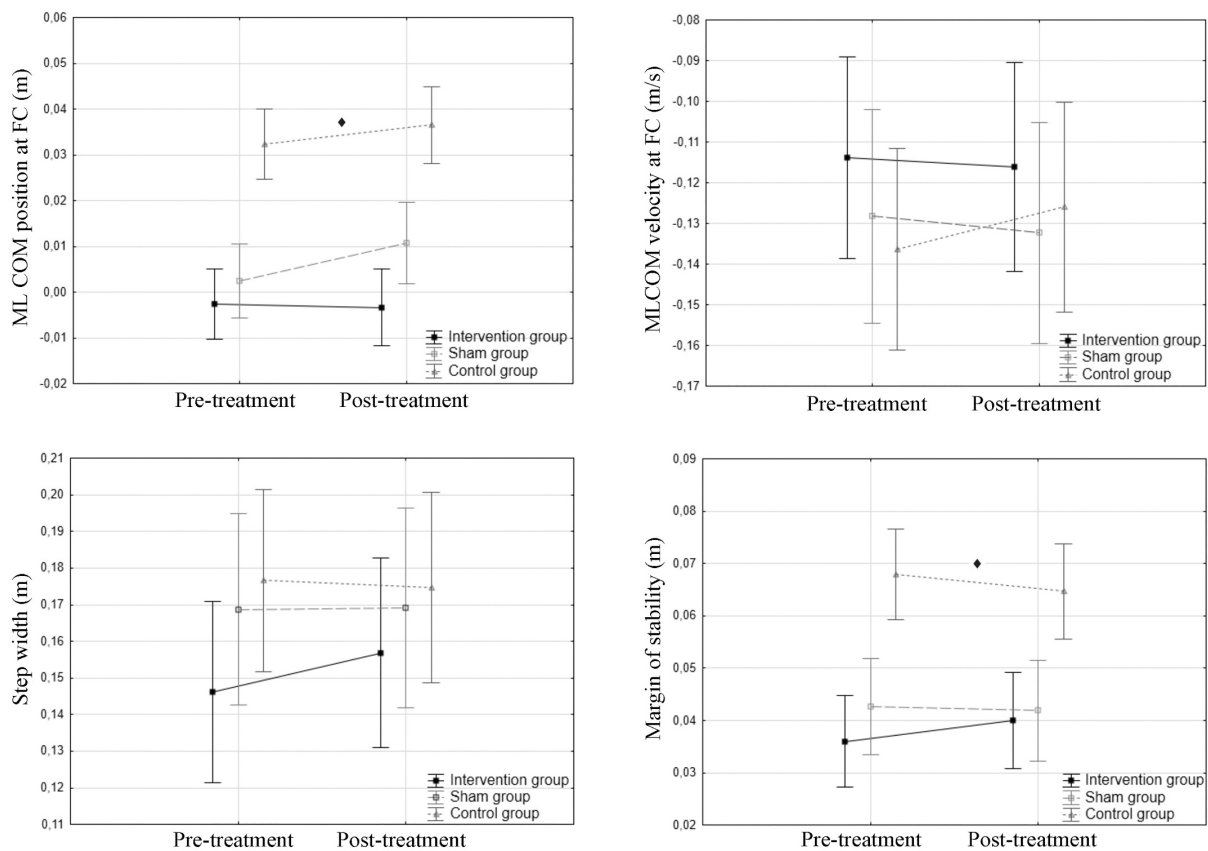


Fig. 5. Effects of group and time on selected stability parameters. COM: center of mass; FC: foot-contact; vertical bars denote 0.95 confidence interval;  $\blacklozenge$  indicates a significant effect of the group as revealed with the ANOVA.

or video game-based exercise program (Maillot et al., 2014) to further improve postural control in PD patients.

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#### Author contributions

TV, AD, IM, GL, SM, PF and EY contributed to project creation, data collection, data analysis, drafted the manuscript, discussed the results, and participated in the revision of the manuscript.

#### Declaration of Competing Interest

None.

#### References

- Bazalgette, D., Zattara, M., Bathien, N., Bouisset, S., Rondot, P., 1987. Postural adjustments associated with rapid voluntary arm movements in patients with Parkinson's disease. *Adv. Neurol.* 45, 371–374.
- Bonora, G., Mancini, M., Carpinella, I., Chiari, L., Horak, F.B., Ferrarin, M., 2017. Gait initiation is impaired in subjects with Parkinson's disease in the OFF state: evidence from the analysis of the anticipatory postural adjustments through wearable inertial sensors. *Gait Posture* 51, 218–221. <https://doi.org/10.1016/j.gaitpost.2016.10.017>.
- Bouisset, S., Do, M.-C., 2008. Posture, dynamic stability, and voluntary movement. *Neurophysiologie Clinique/Clinical Neurophysiology* 38, 345–362. <https://doi.org/10.1016/j.neucli.2008.10.001>.
- Brenière, Y., Do, M.-C., 1986. When and how does steady state gait movement induced from upright posture begin? *J. Biomech.* 19, 1035–1040. [https://doi.org/10.1016/0021-9290\(86\)90120-x](https://doi.org/10.1016/0021-9290(86)90120-x).
- Brenière, Y., Do, M.-C., Bouisset, S., 1987. Are dynamic phenomena prior to stepping essential to walking? *J. Mot. Behav.* 19, 62–76. <https://doi.org/10.1080/00222895.1987.10735400>.
- Caderby, T., Yiou, E., Peyrot, N., Bonazzi, B., Dalleau, G., 2013. Detection of swing heel-off event in gait initiation using force-plate data. *Gait Posture* 37, 463–466. <https://doi.org/10.1016/j.gaitpost.2012.08.011>.
- Caderby, T., Yiou, E., Peyrot, N., de Viviés, X., Bonazzi, B., Dalleau, G., 2017. Effects of changing body weight distribution on Mediolateral stability control during gait initiation. *Front. Hum. Neurosci.* 11, 127. <https://doi.org/10.3389/fnhum.2017.00127>.
- Carlsöö, S., 1966. The initiation of walking. *Acta Anat. (Basel)* 65, 1–9.
- Carpinella, I., Crenna, P., Calabrese, E., Rabuffetti, M., Mazzoleni, P., Nemni, R., Ferrarin, M., 2007. Locomotor function in the early stage of Parkinson's disease. *IEEE Trans Neural Syst Rehabil Eng* 15, 543–551. <https://doi.org/10.1109/TNSRE.2007.908933>.
- Chastan, N., Do, M.-C., Bonneville, F., Tornay, F., Bloch, F., Westby, G.W.M., Dormont, D., Agid, Y., Welter, M.-L., 2009. Gait and balance disorders in Parkinson's disease: impaired active braking of the fall of Centre of gravity. *Mov. Disord.* 24, 188–195. <https://doi.org/10.1002/mds.22269>.
- Crenna, P., Frigo, C., 1991. A motor programme for the initiation of forward-oriented movements in humans. *J. Physiol. Lond.* 437, 635–653. <https://doi.org/10.1113/jphysiol.1991.sp018616>.
- Delafontaine, A., Honeine, J.-L., Do, M.-C., Gagey, O., Chong, R.K., 2015. Comparative gait initiation kinematics between simulated unilateral and bilateral ankle hypomobility: does bilateral constraint improve speed performance? *Neurosci. Lett.* 603, 55–59. <https://doi.org/10.1016/j.neulet.2015.07.016>.
- Delafontaine, A., Gagey, O., Colnaghi, S., Do, M.-C., Honeine, J.-L., 2017. Rigid ankle foot Orthosis deteriorates Mediolateral balance control and vertical braking during gait initiation. *Front. Hum. Neurosci.* 11, 214. <https://doi.org/10.3389/fnhum.2017.00214>.
- Delafontaine, A., Fourcade, P., Honeine, J.L., Ditcharles, S., Yiou, E., 2018. Postural adaptations to unilateral knee joint hypomobility induced by orthosis wear during gait initiation. *Sci. Rep.* 8, 830. <https://doi.org/10.1038/s41598-018-19151-1>.

- Gajdosik, R.L., Vander Linden, D.W., McNair, P.J., Williams, A.K., Riggin, T.J., 2005. Effects of an eight-week stretching program on the passive-elastic properties and function of the calf muscles of older women. *Clin. Biomech.* 20, 973–983. <https://doi.org/10.1016/j.clinbiomech.2005.05.011>.
- Halliday, S.E., Winter, D.A., Frank, J.S., Patla, A.E., Prince, F., 1998. The initiation of gait in young, elderly, and Parkinson's disease subjects. *Gait Posture* 8, 8–14. [https://doi.org/10.1016/S0966-6362\(98\)00020-4](https://doi.org/10.1016/S0966-6362(98)00020-4).
- Hof, A.L., Gazendam, M.G.J., Sinke, W.E., 2005. The condition for dynamic stability. *J. Biomech.* 38, 1–8. <https://doi.org/10.1016/j.jbiomech.2004.03.025>.
- Honeine, J.-L., Schieppati, M., Gagey, O., Do, M.-C., 2013. The functional role of the triceps Surae muscle during human locomotion. *PLoS One* 8. <https://doi.org/10.1371/journal.pone.0052943>.
- Honeine, J.-L., Schieppati, M., Gagey, O., Do, M.-C., 2014. By counteracting gravity, triceps surae sets both kinematics and kinetics of gait. *Phys. Rep.* 2, e00229. <https://doi.org/10.1002/phy.2.229>.
- Honeine, J.-L., Schieppati, M., Crisafulli, O., Do, M.-C., 2016. The neuro-mechanical processes that underlie goal-directed medio-lateral APA during gait initiation. *Front. Hum. Neurosci.* 10. <https://doi.org/10.3389/fnhum.2016.00445>.
- Konor, M.M., Morton, S., Eckerson, J.M., Grindstaff, T.L., 2012. Reliability of three measures of ankle dorsiflexion range of motion. *Int J Sports Phys Ther* 7, 279–287.
- Lepers, R., Brenière, Y., 1995. The role of anticipatory postural adjustments and gravity in gait initiation. *Exp. Brain Res.* 107, 118–124. <https://doi.org/10.1007/bf00228023>.
- Lidstone, S.C., Schulzer, M., Dinelle, K., Mak, E., Sossi, V., Ruth, T.J., de la Fuente-Fernández, R., Phillips, A.G., Stoessl, A.J., 2010. Effects of expectation on placebo-induced dopamine release in Parkinson disease. *Arch. Gen. Psychiatry* 67, 857. <https://doi.org/10.1001/archgenpsychiatry.2010.88>.
- Mancini, M., Zampieri, C., Carlson-Kuhta, P., Chiari, L., Horak, F.B., 2009. Anticipatory postural adjustments prior to step initiation are hypometric in untreated parkinson's disease: an accelerometer-based approach. *Eur. J. Neurol.* 16, 1028–1034. <https://doi.org/10.1111/j.1468-1331.2009.02641.x>.
- Maillot, P., Perrot, A., Hartley, A., Do, M.C., 2014. The braking force in walking: age-related differences and improvement in older adults with exergame training. *J. Aging Phys. Act.* 22 (4), 518–526. <https://doi.org/10.1123/japa.2013-0001>.
- Mancini, M., Nutt, J.G., Horak, F.B., 2020. Chapter 8 - is freezing of gait a balance disorder? In: Mancini, M., Nutt, J.G., Horak, F.B. (Eds.), *Balance Dysfunction in Parkinson's Disease*. Academic Press, pp. 139–161. <https://doi.org/10.1016/B978-0-12-813874-8.00008-8>.
- Martin, R.L., McPoil, T.G., 2005. Reliability of ankle goniometric measurements: a literature review. *J. Am. Podiatr. Med. Assoc.* 95, 564–572. <https://doi.org/10.7547/0950564>.
- Mitchell, S.L., Collins, J.J., De Luca, C.J., Burrows, A., Lipsitz, L.A., 1995. Open-loop and closed-loop postural control mechanisms in Parkinson's disease: increased mediolateral activity during quiet standing. *Neurosci. Lett.* 197, 133–136. [https://doi.org/10.1016/0304-3940\(95\)11924-1](https://doi.org/10.1016/0304-3940(95)11924-1).
- Nakamura, M., Ikezoe, T., Nishishita, S., Umehara, J., Kimura, M., Ichihashi, N., 2017. Acute effects of static stretching on the shear elastic moduli of the medial and lateral gastrocnemius muscles in young and elderly women. *Musculoskelet Sci Pract* 32, 98–103. <https://doi.org/10.1016/j.jmsksp.2017.09.006>.
- Okada, Y., Fukumoto, T., Takatori, K., Nagino, K., Hiraoka, K., 2011a. Abnormalities of the first three steps of gait initiation in patients with Parkinson's disease with freezing of gait. *Parkinson's disease* 2011, 202937. <https://doi.org/10.4061/2011/202937>.
- Okada, Y., Fukumoto, T., Takatori, K., Nagino, K., Hiraoka, K., 2011b. Variable initial swing side and prolonged double limb support represent abnormalities of the first three steps of gait initiation in patients with Parkinson's disease with freezing of gait. *Front. Neurol.* 2, 85. <https://doi.org/10.3389/fneur.2011.00085>.
- Postuma, R.B., Berg, D., Stern, M., Poewe, W., Olanow, C.W., Oertel, W., Obeso, J., Marek, K., Litvan, I., Lang, A.E., Halliday, G., Goetz, C.G., Gasser, T., Dubois, B., Chan, P., Bloem, B.R., Adler, C.H., Deuschl, G., 2015. MDS clinical diagnostic criteria for Parkinson's disease. *Mov. Disord.* 30, 1591–1601. <https://doi.org/10.1002/mds.26424>.
- Raza, C., Anjum, R., Shakeel, N.A., 2019. Parkinson's disease: mechanisms, translational models and management strategies. *Life Sci.* 226, 77–90. <https://doi.org/10.1016/j.lfs.2019.03.057>.
- Shen, X., Wong-Yu, I.S.K., Mak, M.K.Y., 2016. Effects of exercise on falls, balance, and Gait ability in Parkinson's disease: a meta-analysis. *Rev. Neurorehabil Neural Rep.* 30 (6), 512–527. <https://doi.org/10.1177/1545968315613447>.
- Shrier, I., Gossal, K., 2000. Myths and truths of stretching: individualized recommendations for healthy muscles. *Phys. Sportsmed.* 28, 57–63. <https://doi.org/10.3810/psm.2000.08.1159>.
- Vandervoort, A.A., Chesworth, B.M., Cunningham, D.A., Paterson, D.H., Rechner, P.A., Koval, J.J., 1992. Age and sex effects on mobility of the human ankle. *J. Gerontol.* 47, M17–M21. <https://doi.org/10.1093/geronj/47.1.m17>.
- Webber, S.C., Porter, M.M., 2010. Reliability of ankle isometric, isotonic, and isokinetic strength and power testing in older women. *Phys. Ther.* 90, 1165–1175. <https://doi.org/10.2522/ptj.20090394>.
- Winter, D.A., 1995. A.B.C. (anatomy, biomechanics and control) of balance during standing and walking. In: *Waterloo biomechanics*, Waterloo, Ont.
- Winter, D.A., Patla, A.E., Frank, J.S., 1990. Assessment of balance control in humans. *Med Prog Technol* 16, 31–51.
- Yiou, E., Artico, R., Teyssedre, C.A., Labaune, O., Fourcade, P., 2016a. Anticipatory postural control of stability during gait initiation over obstacles of different height and distance made under reaction-time and self-initiated instructions. *Front. Hum. Neurosci.* 10. <https://doi.org/10.3389/fnhum.2016.00449>.
- Yiou, E., Teyssedre, C., Artico, R., Fourcade, P., 2016b. Comparison of base of support size during gait initiation using force-plate and motion-capture system: a bland and Altman analysis. *J. Biomech.* 49, 4168–4172. <https://doi.org/10.1016/j.jbiomech.2016.11.008>.
- Yiou, E., Caderby, T., Delafontaine, A., Fourcade, P., Honeine, J.-L., 2017. Balance control during gait initiation: state-of-the-art and research perspectives. *World J Orthop* 8, 815–828. <https://doi.org/10.5312/wjo.v8.i11.815>.





# **Chapitre 4 : Conclusions et perspectives**

## 4. Conclusions et perspectives

### 4.1. Effets des vibrations corps entier et des étirements musculaires sur la capacité posturo-cinétique

L'objectif général de cette thèse était l'analyse des effets d'une modification de la capacité posturo-cinétique induite par des VCE et des étirements musculaires sur l'organisation biomécanique de la marche et de l'initiation de la marche. Pour réaliser cet objectif, deux études expérimentales et deux revues systématiques de la littérature avec méta-analyses ont été réalisées.

Dans notre **première étude expérimentale**, nous avons montré qu'une séance unique d'application de VCE entraînait une augmentation de la raideur des membres inférieurs au cours de l'initiation de la marche chez les jeunes adultes. Or, cette augmentation de raideur est connue pour être un facteur négatif de la capacité posturo-cinétique (e.g. Hamaoui et al. 2007 ; Bouisset and Lebozec 2002), pouvant impacter négativement la stabilité lors de l'initiation de la marche (Yiou et al. 2016a). Dans notre étude, cette augmentation expérimentale de raideur était cependant surcompensée par une augmentation de l'amplitude des APA sur l'axe médiolatéral, comme en témoignait l'amélioration de la stabilité mesurée par la marge de stabilité. A noter que cette augmentation de la stabilité ne s'accompagnait pas d'une réduction de la performance motrice (mesurée par la vitesse de progression et la longueur du pas), comme il avait pu par ailleurs être montré suite à une hypomobilité expérimentale du membre portant induite par un port d'attelle (Delafontaine et al. 2017). A noter également que la durée des APA n'était pas modifiée par les VCE. Autrement dit, les sujets étaient capables de générer davantage d'APA en un laps de temps équivalent à la condition contrôle. Ces résultats vont donc dans le sens d'une amélioration de l'efficacité de la phase posturale des sujets jeunes après une séance de VCE. A l'inverse, chez les sujets âgés en bonne santé, les VCE n'ont eu d'effet immédiat sur aucune des

variables biomécaniques analysées. La raideur du membre portant n'ayant pas augmenté dans cette population après les VCE, on peut penser qu'aucune adaptation posturale n'ait été nécessaire pour assurer la stabilité posturale.

Sur la base de ces observations, nous avons proposé que l'augmentation de raideur du membre portant est prise en compte par le système nerveux central lors de la programmation des APA. La raideur étant perçue comme une menace pour l'équilibre, cette contrainte est anticipée par le système nerveux central. Ce dernier met alors en place une adaptation posturale contre-perturbatrice pour préserver la stabilité du corps. Ce résultat est en accord avec un corpus d'études récentes montrant que, chez le jeune adulte, le système nerveux central adapte systématiquement les mécanismes de contrôle de l'équilibre (en particulier les APA, la largeur du pas et/ou la vitesse de progression) en situation de contrainte posturale de façon à maintenir une stabilité optimale. C'est ce qui a été montré, par exemple, lorsque le contrôle de la stabilité posturale est rendu plus difficile par l'ajout d'une contrainte *externe*, telle que la présence d'un obstacle à enjamber (Yiou et al. 2016a), l'augmentation de la vitesse de progression (Caderby et al. 2014), l'augmentation du niveau de pression temporelle imposé par l'expérimentateur (Yiou et al. 2016b), ou encore l'ajout d'une masse au niveau des hanches des participants (Caderby et al. 2014, 2017). Nos résultats permettent ainsi d'élargir nos connaissances sur la capacité d'adaptation du système nerveux en présence d'une contrainte posturale *interne*.

Cette modulation adaptative des mécanismes du contrôle postural partage des caractéristiques semblables à la régulation de l'*homéostasie* que l'on retrouve dans les mécanismes physiologiques (Yiou et al. 2017). En effet, l'homéostasie correspond à « *la stabilisation active des variables du corps (température, glycémie, pression artérielle etc...), en réponse à tout stimulus qui tendrait à perturber son état ou sa fonction normale* » (Cooper 2008). Appliqué au cas de l'initiation de la marche, le système nerveux central adapterait les mécanismes de contrôle de la stabilité posturale de façon à maintenir celle-ci inchangée, voire à l'améliorer dans le cas de la présente étude. Ce dernier pourrait ainsi avoir *internalisé* des valeurs de référence de stabilité et

s'efforceraient de les maintenir inchangées afin de préserver l'intégrité physique du corps malgré les différentes sources de perturbation. La marge de stabilité pourrait correspondre à un tel type de variable régulée.

Dans notre **seconde étude expérimentale**, la capacité posturo-cinétique était manipulée par le biais d'étirements musculaires supposés améliorer la mobilité de la chaîne posturale (la mobilité des chevilles en l'occurrence) chez le sujet atteint de la maladie de Parkinson. Cette population est en effet connue pour avoir une mobilité réduite de la chaîne posturale, ce qui pourrait constituer un frein à leur performance posturale. Comme pour la raideur, la mobilité de la chaîne posturale est en effet également connue pour être un facteur clef de la capacité posturo-cinétique (Le Bozec and Bouisset 2004 ; Diakhaté et al. 2013 ; Delafontaine et al. 2015, 2017, 2018 ; Alamini-Rodrigues and Hamaoui 2017 ; Hamaoui and Alamini-Rodrigues 2017). Nos résultats ont montré que, chez ces patients, une seule séance d'étirements des triceps suraux permettait d'augmenter de façon substantielle la mobilité des chevilles. Ce gain de mobilité n'avait cependant aucun effet sur la stabilité posturale au cours de l'initiation de la marche. En revanche, la vitesse de progression antéropostérieure du centre des masses au décolllement du talon (i.e. pendant la phase des APA) était augmentée. Comme pour les VCE, la durée des APA restait inchangée après l'application du traitement. Il semblerait donc que l'amplitude des APA soit plus sensible que la durée des APA à ces deux types d'entraînement.

Les patients ayant reçu le traitement d'étirements du triceps sural à court terme ont donc, au final, développé davantage de forces propulsives vers l'avant durant la phase posturale, avec une durée équivalente. A nouveau, ce résultat montre une amélioration de l'efficacité de la phase posturale des sujets qui ont reçu le traitement. En conséquence directe, la vitesse de marche moyenne et le pic de vitesse à la fin du premier pas, c'est-à-dire la performance motrice, ont significativement augmenté. Ce résultat nous semble important car la difficulté à générer des APA d'intensité convenable ainsi que la lenteur des mouvements font partie des symptômes majeurs

de ces patients (Bazalgette et al. 1987 ; Halliday et al. 1998 ; Mancini et al. 2009 ; Okada et al. 2011a, b ; Bonora et al. 2017).

Au final, il ressort de ces deux études expérimentales que le concept de capacité posturo-cinétique fournit un cadre théorique pertinent pour investiguer la problématique de l'adaptabilité du système nerveux à des contraintes posturales, mais également pour cibler des protocoles d'amélioration de la fonction posturale dynamique, notamment chez le sujet atteint de la maladie de Parkinson. Combinés à nos deux études systématiques, l'ensemble de nos travaux permettent d'envisager des applications cliniques que nous développons dans le paragraphe ci-dessous.

## **4.2. Applications pratiques**

Maintenir une activité physique régulière est essentielle pour la santé. Elle réduit l'ensemble des risques de mortalité toutes causes confondues, notamment le risque de chute chez les personnes âgées. L'inactivité physique reste malgré tout trop répandue dans cette population, encore plus chez les patients atteints de maladies chroniques, et les personnes âgées devraient être aussi actives que leurs capacités fonctionnelles le permettent (Chodzko-Zajko et al. 2009 ; Lee et al. 2017 ; Mora and Valencia 2018). La planification des exercices physiques et les recommandations aux patients doivent tenir compte des besoins et des possibilités physiques et mentales des personnes, de façon que les bénéfices soient perceptibles et supérieurs aux efforts nécessaires pour s'exercer. Les VCE et les étirements permettent de répondre à ces problématiques en améliorant les capacités fonctionnelles des patients selon certaines conditions. L'intégration de ces méthodes d'entraînement peut se faire de manière raisonnée, à partir des travaux de cette thèse.

Notre étude expérimentale suggère que les VCE peuvent améliorer l'équilibre lors de l'initiation de la marche à court terme, à condition que les participants soient en mesure de développer des stratégies adaptatives par le biais des différents

mécanismes du contrôle postural afin de compenser l'augmentation de la raideur lorsqu'elle survient. Dans notre étude, cela a été observé chez les jeunes adultes sains, mais pas chez les personnes âgées en bonne santé. Dans cette population, les VCE ne semblent pas avoir d'effets immédiats sur les paramètres biomécaniques de l'initiation de la marche. Cette méthode est donc davantage indiquée pour les programmes de rééducation de plusieurs semaines pour lesquels il a été montré qu'elle permet d'améliorer la force, l'endurance, la masse musculaire et l'équilibre (Orr 2015 ; Park et al. 2015 ; Rogan et al. 2017 ; Lam et al. 2018). Notre méta-analyse a montré que la performance au *Timed up and go test* et au *10-meter walk test* s'améliore également avec ce type d'entraînement (Fischer al. 2019), ce qui en fait un outil de rééducation important dans le maintien de l'autonomie de ces patients. De plus, il est à noter que Les VCE sont bien tolérées par l'ensemble des patients (moins de 15% d'abandon dans la majorité des études, Fischer et al. 2019).

Nous avons également montré qu'une session de VCE avec les paramètres d'entraînement de notre étude expérimentale (4 séries de 45s, 1 minute de repos entre les séries, fréquence de 50Hz, 2mm de déplacement vertical) n'altère pas l'équilibre des personnes âgées en bonne santé. Il s'agit donc à priori d'une méthode d'entraînement sécuritaire, d'autant que les plateformes vibrantes offrent généralement des points d'appui et permettent de travailler en position statique et/ou assise. Cependant, notre étude expérimentale a aussi mis en évidence que les VCE peuvent potentiellement augmenter la raideur de la jambe portante lors de l'initiation de la marche (e.g. chez les sujets jeunes). Chez les personnes âgées, cette raideur pourrait survenir avec des paramètres différents ou des durées d'exposition plus longues (Han et al. 2017). Une possible augmentation de la raideur est donc à surveiller après une séance de VCE, car elle pourrait augmenter le risque de chute chez des sujets incapables de mettre en place les adaptations posturales contre-perturbatrices nécessaires au maintien de l'équilibre.

Les étirements améliorent l'amplitude des mouvements immédiatement après une séance et après plusieurs semaines d'entraînement (Radford et al. 2006). Ils sont

donc indiqués, d'une part, en sport lorsque la performance de l'activité exige des mouvements de grande amplitude et, d'autre part, en rééducation dans toute situation où la raideur augmente (vieillesse, peur de tomber, sédentarité, maladies neuromusculaires etc., Portero et al. 2015). Or, nous avons montré dans notre étude expérimentale que les patients parkinsoniens ont une raideur plus importante autour de la cheville par rapport à des sujets sains du même âge. Ce résultat est en adéquation avec la littérature scientifique où l'on retrouve régulièrement la rigidité comme symptôme principal de la maladie (Lee and Gilbert 2016 ; Tysnes and Storstein 2017; Raza et al. 2019), notamment autour de la cheville (Wu et al. 2020). Immédiatement après une séance, les étirements du triceps sural améliorent non seulement l'amplitude passive en dorsiflexion, mais aussi la vitesse de marche des patients. Ce traitement semble donc efficace pour faciliter le déclenchement du premier pas de ces patients, sans altération de la stabilité, même si l'effet placebo n'est pas à exclure. D'autres études sont nécessaires pour confirmer ces résultats, notamment chez des patients qui sont sujets au « freezing », et pour évaluer les effets à long terme des étirements musculaires. Notre revue de littérature avec méta-analyse montre qu'il n'est pour le moment pas possible d'affirmer que les étirements à eux seuls suffisent à améliorer les capacités de marche des patients sur le long terme, malgré quelques résultats positifs isolés (Gajdosik et al. 2005 ; Cristopoliski et al. 2009 ; Hotta et al. 2019). A partir de ces données, nous recommandons donc d'intégrer les étirements dans des programmes d'exercices physiques complets, avec notamment du renforcement musculaire, du travail de l'endurance et de l'équilibre (Lee et al. 2017 ; Mora and Valencia 2018). Dans cette optique, rappelons que l'entraînement par VCE apparaît aussi comme un outil complémentaire intéressant car il permet d'améliorer l'ensemble de ces capacités physiques (Smith et al. 2016), même si des activités plus fonctionnelles et en charge (la marche, la montée et descente des escaliers, se lever d'une chaise etc.) sont à préconiser pour maintenir l'autonomie des patients sur le plan locomoteur.

Enfin, notons aussi que la prise en charge précoce et pluridisciplinaire des patients en rééducation pourrait permettre de ralentir la progression de la maladie de



Parkinson et limiter le recours aux traitements médicamenteux (Frazzitta et al. 2015). Proposer des activités physiques adaptées aux capacités des patients dès le début de la prise en charge permettrait ainsi d'améliorer les capacités fonctionnelles des patients à court et long termes (Martínez-Velilla et al. 2016). Améliorer précocement l'autonomie des patients apparaît évidemment important pour la santé des patients, mais aussi pour soulager le travail des équipes soignantes et réduire les coûts liés aux hospitalisations de longue durée. L'amélioration des capacités physiques des patients est un moyen de favoriser leur participation aux soins et inversement de réduire la dépendance aux soignants ou aux aides matérielles parfois coûteuses (Martínez-Velilla et al. 2016). Les exercices étudiés ici, VCE et étirements, permettent de répondre à ces problématiques. Ils sont peu coûteux, bien tolérés et réalisables précocement, même chez des patients avec des incapacités physiques importantes.

### **4.3. Perspectives de recherche**

Les résultats de notre étude expérimentale sur les VCE ont montré que cet exercice est susceptible de modifier la capacité posturo-cinétique des sujets en augmentant la raideur des membres inférieurs. Notre programme n'a cependant eu aucun impact sur la raideur des sujets âgés, mais il est possible que la modification de certains paramètres de l'entraînement apporte de nouveaux résultats. En effet, il a été montré que le type de vibrations (i.e. synchrone ou asynchrone), la fréquence ou la position des sujets peuvent modifier l'activité électromyographique lors d'une session de VCE (Ritzmann et al. 2013). En augmentant la fréquence des vibrations par exemple, il est possible que la raideur des membres inférieurs augmente aussi chez des sujets âgés. Il serait alors intéressant d'observer si ces derniers sont capables de moduler les APA afin de maintenir une stabilité optimale lors de l'initiation de la marche.

Les étirements du triceps sural ont permis d'améliorer l'amplitude en dorsiflexion de la cheville et d'améliorer la capacité posturo-cinétique des patients parkinsoniens à court terme. Un effet placebo n'étant pas à exclure, une étude

randomisée est actuellement en cours pour étudier les effets d'un programme d'étirements du triceps sural sur 4 semaines, afin de vérifier si ces effets perdurent à long terme. Les résultats de cette étude permettront de clarifier l'intérêt des étirements dans les programmes de rééducation de ces patients, car à ce jour aucune donnée n'a été retrouvée dans la littérature pour cette population (Vialleron et al. 2020), bien que les étirements soient recommandés pour lutter contre la raideur (Abbruzzese et al. 2016).

Enfin, l'entraînement de la marche avec enjambement d'obstacles nous apparaît aussi comme une piste de recherche à explorer pour améliorer les capacités posturales des patients hémiparétiques ou parkinsoniens. En effet, des exercices d'enjambement d'obstacles sont régulièrement utilisés en rééducation pour provoquer un allongement du temps d'appui unipodal chez les patients présentant une diminution de la phase portante unipodale lors de la marche, associée à un déficit de hauteur et de longueur du pas (Coquisart 2012). Lors de l'initiation de la marche avec obstacle, la durée de la phase oscillante augmente avec l'augmentation de la hauteur de l'obstacle, prolongeant ainsi la phase portante (Chou et al. 2001 ; Hahn and Chou 2004 ; Yiou et al. 2016a). Il a été montré que l'augmentation de la hauteur de l'obstacle favorise aussi l'augmentation de l'amplitude des APA sur l'axe médiolatéral (Yiou et al. 2016b). Des résultats similaires ont été observés chez les jeunes adultes comme chez les personnes âgées (Chou et al. 2001 ; Hahn and Chou 2004). Les patients pourraient ainsi apprendre à générer de nouveau des APA efficaces sur l'axe médiolatéral afin de mieux transférer le poids du corps vers la jambe portante et maintenir un équilibre dynamique satisfaisant lors de la phase d'exécution. Ce type d'entraînement peut avoir son importance chez des patients présentant une diminution des APA médiolatéraux et une dégradation de la stabilité latérale, comme nous l'avons montré chez les patients parkinsoniens. Cet exercice pourrait aussi être utilisé chez les patients hémiparétiques qui ont des difficultés à transférer latéralement le poids du corps vers la jambe portante (Delafontaine et al. 2019b).

Nous avons montré au travers d'une revue systématique de la littérature (Annexe 1, Delafontaine et al. 2019b) que les patients hémiplegiques présentent une diminution de l'amplitude des APA sur les deux axes, antéropostérieur et médiolatéral, ainsi qu'une augmentation de la durée de la phase posturale. De plus, les patients ont plus de difficultés à initier la marche avec la jambe saine comme jambe oscillante, car la faiblesse musculaire du côté atteint entraîne des difficultés à supporter le poids du corps lors de la phase unipodale. Dans cette optique, des programmes de renforcement musculaire spécifiques du membre inférieur atteint pourraient améliorer certaines composantes de l'initiation de la marche. Il serait par exemple intéressant d'étudier si le renforcement des abducteurs de hanches permet d'améliorer les déplacements médiolatéraux du centre des pressions et du centre des masses lors des APA. Le renforcement des muscles extenseurs (e.g. grand fessier, quadriceps, triceps sural) pourrait améliorer les forces propulsives mais aussi le temps d'appui unipodal. Enfin, le renforcement du tibial antérieur est susceptible d'améliorer l'amplitude des APA dans le plan sagittal.

L'ensemble de ces perspectives de recherche, dans la lignée des travaux de cette thèse, permettront de mieux adapter les programmes de rééducation dont les objectifs sont de réduire le risque de chute et améliorer l'autonomie des patients sur le plan locomoteur.

# Références

## Références

Cette section regroupe les références citées dans les chapitres 1, 2 et 4.

- Abbruzzese G, Marchese R, Avanzino L, Pelosin E (2016). Rehabilitation for Parkinson's disease: Current outlook and future challenges. *Parkinsonism Relat Disord* 1:S60-4. doi: 10.1016/j.parkreldis.2015.09.005
- Adkin AL, Frank JS, Carpenter MG, Peysar GW (2002). Fear of falling modifies anticipatory postural control. *Exp Brain Res* 143(2):160-70. doi: 10.1007/s00221-001-0974-8
- Agid Y. (1990). From posture to initiation of movement *Rev Neurol (Paris)*;146(10):536-42.
- André-Thomas (1940). *Equilibre et équilibration*. Masson (Eds.), Paris.
- Alamini-Rodrigues C, Hamaoui A (2017). Effect of three different lumbar splints on posturokinetic capacity when performing the sit-to-stand task. *Ann Phys Rehabil Med* 60:406–409. <https://doi.org/10.1016/j.rehab.2016.09.003>
- Alexander NB, Shepard N, Gu MJ, Schultz A (1992). Postural control in young and elderly adults when stance is perturbed: kinematics. *J. Gerontol* 47(3):M79-87. doi:10.1093/geronj/47.3.m79.
- Alfuth M, Beiring A, Klein D, Rosenbaum D (2012). Acute effects of whole body vibration on foot sole sensitivity and plantar pressures during gait initiation. *J Foot Ankle Res* 5:O24. <https://doi.org/10.1186/1757-1146-5-S1-O24>
- Aruin AS (2016). Enhancing Anticipatory Postural Adjustments: A Novel Approach to Balance Rehabilitation. *J Nov Physiother* 6: <https://doi.org/10.4172/2165-7025.1000e144>
- Aruin AS, Forrest WR, Latash ML (1998). Anticipatory postural adjustments in conditions of postural instability. *Electroencephalogr Clin Neurophysiol* 109:350–359
- Aruin AS, Ganesan M, Lee Y (2017). Improvement of postural control in individuals with multiple sclerosis after a single-session of ball throwing exercise. *Multiple Sclerosis and Related Disorders* 17:224–229. <https://doi.org/10.1016/j.msard.2017.08.013>
- Aruin AS, Kanekar N, Lee Y-J, Ganesan M (2015). Enhancement of anticipatory postural adjustments in older adults as a result of a single session of ball throwing exercise. *Exp Brain Res* 233:649–655. <https://doi.org/10.1007/s00221-014-4144-1>

- Aruin AS, Latash ML (1995). The role of motor action in anticipatory postural adjustments studied with self-induced and externally triggered perturbations. *Exp Brain Res* 106:291–300. <https://doi.org/10.1007/BF00241125>
- Babinski J. De l'asynergie cérébelleuse (1899). *Revue Neurologique*. 7:806-816.
- Bauby CE, Kuo AD (2000). Active control of lateral balance in human walking. *J Biomech* 33:1433–1440. [https://doi.org/10.1016/s0021-9290\(00\)00101-9](https://doi.org/10.1016/s0021-9290(00)00101-9)
- Bazalgette D, Zattara M, Bathien N, Bouisset S, Rondot P (1987). Postural adjustments associated with rapid voluntary arm movements in patients with Parkinson's disease. *Adv Neurol* 45:371–374
- Bear MF, Connors BW, Paradiso MA (2015). *Neurosciences, à la découverte du cerveau*, Pradel. Wolters Kluwer, USA
- Belen'kiï VE, Gurfinkel' VS, Pal'tsev EI (1967). [Control elements of voluntary movements]. *Biofizika* 12:135–141
- Benvenuti F, Stanhope SJ, Thomas SL, Panzer VP, Hallett M (1995). Postural adjustments associated with fast elbow flexion movements in self-paced and reaction time conditions. *Electroencephalography and Clinical Neurophysiology* 3:48P
- Bohannon RW (1997). Comfortable and maximum walking speed of adults aged 20-79 years: reference values and determinant. *Age Ageing*. 26(1):15-9. doi:10.1093/ageing/26.1.15.
- Bolzoni F, Bruttini C, Esposti R, Castellani C, Cavallari P(2015). Transcranial direct current stimulation of SMA modulates anticipatory postural adjustments without affecting the primary movement. *Behavioural Brain Research* 291:407-413 doi:10.1016/j.bbr.2015.05.044 PMID: 26055201
- Bolzoni F, Esposti R, Marchese S, Pozzi N, Ramirez-Pasos U, Isaias I, Cavallari P (2018). Disrupt of Intra-Limb APA Pattern in Parkinsonian Patients Performing Index-Finger Flexion. *Front Physiol* 9:1745. doi: 10.3389/fphys.2018.01745
- Bonnetblanc F, Martin O, Teasdale N (2004). Pointing to a target from an upright standing position: anticipatory postural adjustments are modulated by the size of the target in humans. *Neuroscience Letters* 358(3), 181-184. <https://doi.org/10.1016/j.neulet.2004.01.020>
- Bonora G, Mancini M, Carpinella I, Chiari L, Horak FB, Ferrarin M (2017). Gait initiation is impaired in subjects with Parkinson's disease in the OFF state: Evidence from the analysis of the anticipatory postural adjustments through wearable inertial sensors. *Gait Posture* 51:218–221. <https://doi.org/10.1016/j.gaitpost.2016.10.017>

- Bouisset S, Do MC (2008). Posture, dynamic stability, and voluntary movement. *Neurophysiologie Clinique/Clinical Neurophysiology* 38:345–362. <https://doi.org/10.1016/j.neucli.2008.10.001>
- Bouisset S, Duchêne JL (1994). Is body balance more perturbed by respiration in seating than in standing posture? *Neuroreport* 5:957–960. <https://doi.org/10.1097/00001756-199404000-00026>
- Bouisset S, Le Bozec S (2002). Bouisset, S. and Le Bozec, S. (2002) Posturo-kinetic capacity and postural function in voluntary movements. In Latash, ML (Ed), *Progress in Motor Control, volume II: Structure-Function Relations in Voluntary Movements*. Human Kinetics. Chapter 3: 25-52. pp 25–52
- Bouisset S, Maton B. *Muscles, Posture et mouvement*. Hermann Paris, 1995
- Bouisset S, Maton B (1996). *Muscles, posture et mouvement: bases et applications de la méthode électromyographique*. Hermann, Paris
- Bouisset S, Richardson J, Zattara M (2000). Do anticipatory postural adjustments occurring in different segments of the postural chain follow the same organisational rule for different task movement velocities, independently of the inertial load value? *Exp Brain Res* 132:79–86. <https://doi.org/10.1007/s002219900228>
- Bouisset S, Zattara M (1981). A sequence of postural movements precedes voluntary movement. *Neuroscience Letters* 22:263–270. [https://doi.org/10.1016/0304-3940\(81\)90117-8](https://doi.org/10.1016/0304-3940(81)90117-8)
- Bouisset S, Zattara M (1987). Biomechanical study of the programming of anticipatory postural adjustments associated with voluntary movement. *J Biomech* 20:735–742
- Borelli GA. *De Motu Animalium* (Traduit par Maquet P, 1989). Springer Verlag, Berlin, 1680.
- Boyer KA, Johnson RT, Banks JJ, Jewell C, Hafer JF (2017). Systematic review and meta-analysis of gait mechanics in young and older adults. *Experimental Gerontology* 95:63–70. <https://doi.org/10.1016/j.exger.2017.05.005>
- Brenière Y, Bril B (1998). Development of postural control of gravity forces in children during the first 5 years of walking. *Exp Brain Res* 121:255–262. <https://doi.org/10.1007/s002210050458>
- Brenière Y, Do MC (1991). Control of gait initiation. *J Mot Behav* 23:235–240. <https://doi.org/10.1080/00222895.1991.9942034>

- Brenière Y, Do MC, Bouisset S (1987). Are dynamic phenomena prior to stepping essential to walking? *J Mot Behav* 19:62–76. <https://doi.org/10.1080/00222895.1987.10735400>
- Bril B, Breniere Y (1992). Postural requirements and progression velocity in young walkers. *J Mot Behav* 24:105–116. <https://doi.org/10.1080/00222895.1992.9941606>
- Bruttini C, Esposti R, Bolzoni F, Cavallari P (2016). Higher Precision in Pointing Movements of the Preferred vs. Non-Preferred Hand Is Associated with an Earlier Occurrence of Anticipatory Postural Adjustments. *Front. Hum. Neurosci.*, 18 July 2016 | <https://doi.org/10.3389/fnhum.2016.00365>
- Bryant MS, Rintala DH, Hou JG, Charness AL, Fernandez AL, Collins RL, Baker J, Lai EC, Protas EJ (2011). Gait variability in Parkinson's disease: influence of walking speed and dopaminergic treatment. *Neurol Res* 33(9):959-64. doi: 10.1179/1743132811Y.0000000044
- Burke D, Hagbarth KE, Löfstedt L, Wallin BG (1976). The responses of human muscle spindle endings to vibration during isometric contraction. *J Physiol* 261:695–711
- Burleigh-Jacobs A, Horak FB, Nutt JG, Obeso JA (1997). Step initiation in Parkinson's disease: influence of levodopa and external sensory triggers. *Mov Disord* 12:206–215. <https://doi.org/10.1002/mds.870120211>
- Caderby T, Dalleau G, Leroyer P, Bonazzi B, Chane-Teng D, Do MC (2013). Does an additional load modify the Anticipatory Postural Adjustments in gait initiation? *Gait Posture* 37(1):144-6. doi:10.1016/j.gaitpost.2012.06.012.
- Caderby T, Yiou E, Peyrot N, Begon M, Dalleau G (2014). Influence of gait speed on the control of mediolateral dynamic stability during gait initiation. *J Biomech* 47:417–423. <https://doi.org/10.1016/j.jbiomech.2013.11.011>
- Caderby T, Yiou E, Peyrot N, de Viviés X, Bonazzi B, Dalleau G (2017). Effects of Changing Body Weight Distribution on Mediolateral Stability Control during Gait Initiation. *Front Hum Neurosci* 11:127. <https://doi.org/10.3389/fnhum.2017.00127>
- Caliandro P, Ferrarin M, Cioni M, Bentivoglio A R, Minciotti I, D'Urso P I, Tonali P A, Padua L (2011). Levodopa effect on electromyographic activation patterns of tibialis anterior muscle during walking in Parkinson's disease. *Gait & Posture* 33:436–441. <https://doi.org/10.1016/j.gaitpost.2010.12.020>
- Carlsöö S (1966). The initiation of walking. *Acta Anat (Basel)* 65:1–9
- Caronni A, Bolzoni F, Esposti R, Bruttini C, Cavallari P (2013). Accuracy of pointing movements relies upon a specific tuning between anticipatory postural



- adjustments and prime mover activation. *Acta Physiol (Oxf)* 208(1):111-24. doi:10.1111/apha.12081. Epub 2013 Mar 11.
- Chapman GJ, Scally A, Buckley JG (2012). Importance of binocular vision in foot placement accuracy when stepping onto a floor-based target during gait initiation. *Experimental Brain Research* 216:71-80. <https://doi.org/10.1007/s00221-011-2910-x>
- Chastan N, Debono B, Maltete D, Weber J (2008). Discordance between measured postural instability and absence of clinical symptoms in Parkinson's disease patients in the early stages of the disease. *Mov. Disord.* 23:366-372. <https://doi.org/10.1002/mds.21840>
- Chastan N, Do MC, Bonneville F, Torny F, Bloch F, Westby GWM, Dormont D, Agid Y, Welter ML (2009a). Gait and balance disorders in Parkinson's disease: impaired active braking of the fall of centre of gravity. *Mov Disord* 24:188-195. <https://doi.org/10.1002/mds.22269>
- Chastan N, Welter ML, Do MC, Tézenas du Montcel S, Agid T (2006). Control of vertical components of gait during initiation of walking in normal adults. *International Congress on Gait and Mental Function, Madrid.*
- Chastan N, Westby GWM, Yelnik J, Bardinet E, Do MC, Agid Y, Welter ML (2009b). Effects of nigral stimulation on locomotion and postural stability in patients with Parkinson's disease, *Brain* 132:172-184.
- Chastan N, Westby GWM, du Montcel ST, Do MC, Chong R.K, Agid Y, Welter ML (2010). Influence of sensory inputs and motor demands on the control of the centre of mass velocity during gait initiation in humans. *Neuroscience Letters* 469:400-404. <https://doi.org/10.1016/j.neulet.2009.12.038>
- Chaumette T, Lebouvier T, Coron E, Damier P, Derkinderen P, Neunlist M, Bruley des Varannes S (2009). Maladie de Parkinson : des atteintes spécifiques du système nerveux entérique. *Gastroentérologie Clinique et Biologique* 33: A13 [https://doi.org/10.1016/S0399-8320\(09\)72621-1](https://doi.org/10.1016/S0399-8320(09)72621-1)
- Chodzko-Zajko WJ, Proctor DN, Fiatarone Singh MA, Minson CT, Nigg CR, Salem GJ, Skinner JS (2009). American College of Sports Medicine position stand. Exercise and physical activity for older adults. *Med Sci Sports Exerc* 41(7):1510-30. doi: 10.1249/MSS.0b013e3181a0c95c
- Chong RKY, Chastan N, Welter M-L, Do MC (2009). Age-related changes in the center of mass velocity control during walking. *Neuroscience Letters* 458:23-27. <https://doi.org/10.1016/j.neulet.2009.04.022>

- Chou LS, Kaufman KR, Brey RH, Draganich LF (2001). Motion of the whole body's center of mass when stepping over obstacles of different heights. *Gait Posture* 13:17–26
- Collins SH, Kuo AD (2013). Two independent contributions to step variability during over-ground human walking. *PLoS ONE* 8:e73597. <https://doi.org/10.1371/journal.pone.0073597>
- Cooper SJ (2008). From Claude Bernard to Walter Cannon. Emergence of the concept of homeostasis. *Appetite* 51:419–427. <https://doi.org/10.1016/j.appet.2008.06.005>
- Coquisart L (2012). Approche gériatrique de la proprioceptivité. *Kinésithérapie, la Revue* 12:72–76. <https://doi.org/10.1016/j.kine.2012.07.013>
- Corrêa PS, Pagnussat AS, Cabeleira MEP, Schifino GP, de Mello Rieder CR, da Silva Junior N, Cechetti F (2019). Is the dopaminergic loss associated with gait and postural impairments in subjects with Parkinson's disease at different motor stages ? *Eur J Neurosci* 50(12):3889–3895. doi: 10.1111/ejn.14522
- Cordo PJ, Nashner LM (1982). Properties of postural adjustments associated with rapid arm movements. *J Neurophysiol* 47:287–302. <https://doi.org/10.1152/jn.1982.47.2.287>
- Couillandre A, Brenière Y (2003). How Does the Heel-Off Posture Modify Gait Initiation Parameter Programming? *Journal of Motor Behavior* 35:221–227. <https://doi.org/10.1080/00222890309602136>
- Couillandre A, Maton B, Brenière Y (2002). Voluntary toe-walking gait initiation: electromyographical and biomechanical aspects. *Exp Brain Res* 147:313–321. <https://doi.org/10.1007/s00221-002-1254-y>
- Crenna P, Frigo C (1991). A motor programme for the initiation of forward-oriented movements in humans. *J Physiol (Lond)* 437:635–653. <https://doi.org/10.1113/jphysiol.1991.sp018616>
- Crenna P, Frigo C, Massion J, Pedotti A (1987). Forward and backward axial synergies in man. *Exp Brain Res* 65:538–548. <https://doi.org/10.1007/BF00235977>
- Cristopoliski F, Barela JA, Leite N, Fowler NE, Felix Rodacki AL (2009). Stretching Exercise Program Improves Gait in the Elderly. *Gerontology* 55:614–620. <https://doi.org/10.1159/000235863>
- Cruz-Jimenez M (2017). Normal Changes in Gait and Mobility Problems in the Elderly. *Phys Med Rehabil Clin N Am* 28:713–725. <https://doi.org/10.1016/j.pmr.2017.06.005>

- Daoud AI, Geissler GJ, Wang F, Saretsky J, Daoud YA, Lieberman DE (2012). Foot strike and injury rates in endurance runners: a retrospective study. *Med Sci Sports Exerc* 44:1325–1334. <https://doi.org/10.1249/MSS.0b013e3182465115>
- Day, B. L., Steiger, M. J., Thompson, P. D., & Marsden, C. D. (1993).. Effect of vision and stance width on human body motion when standing: implications for afferent control of lateral sway. *The Journal of Physiology*, 469, 479-499.
- Dean JC, Kautz SA (2015). Foot placement control and gait instability among people with stroke. *J Rehabil Res Dev* 52:577–590. <https://doi.org/10.1682/JRRD.2014.09.0207>
- Debove L, Bru N, Couderc M, Noé F, Paillard T (2017). Physical activity limits the effects of age and Alzheimer’s disease on postural control. *Neurophysiol Clin* 47:301–304. <https://doi.org/10.1016/j.neucli.2017.03.005>
- Delafontaine A, Fourcade P, Honeine JL, Ditcharles S, Yiou E (2018). Postural adaptations to unilateral knee joint hypomobility induced by orthosis wear during gait initiation. *Sci Rep* 8:830. <https://doi.org/10.1038/s41598-018-19151-1>
- Delafontaine A, Gagey O, Colnaghi S, Do MC, Honeine JL (2017). Rigid Ankle Foot Orthosis Deteriorates Mediolateral Balance Control and Vertical Braking during Gait Initiation. *Front Hum Neurosci* 11:214. <https://doi.org/10.3389/fnhum.2017.00214>
- Delafontaine A, Honeine JL, Do MC, Gagey O, Chong RK (2015). Comparative gait initiation kinematics between simulated unilateral and bilateral ankle hypomobility: Does bilateral constraint improve speed performance? *Neurosci Lett* 603:55–59. <https://doi.org/10.1016/j.neulet.2015.07.016>
- Delafontaine A, Vialleron T, Fischer M, Laffaye G, Chèze L, Artico R, Genêt F, Fourcade P, Yiou E (2019a). Acute Effects of Whole-Body Vibration on the Postural Organization of Gait Initiation in Young Adults and Elderly: A Randomized Sham Intervention Study. *Frontiers in Neurology* 10:1023. <https://doi.org/10.3389/fneur.2019.01023>
- Delafontaine A, Vialleron T, Hussein T, Yiou E, Honeine JL, Colnaghi S (2019b). Anticipatory Postural Adjustments During Gait Initiation in Stroke Patients. *Frontiers in Neurology* 10:352. <https://doi.org/10.3389/fneur.2019.00352>
- Delval A, Krystkowiak P, Blatt JL, Labyt E, Destée A, Derambure P, Defebvre L (2005). Caractérisation des ajustements posturaux lors d’une initiation de la marche déclenchée par un stimulus sonore et autocommandée chez 20 sujets sains. *Neurophysiologie Clinique/Clinical Neurophysiology* 35:180–190. <https://doi.org/10.1016/j.neucli.2006.01.002>

- Delval A, Tard C, Defebvre L (2014). Why we should study gait initiation in Parkinson's disease. *Neurophysiol Clin* 44(1):69-76. doi: 10.1016/j.neucli.2013.10.127
- Demain A, Westby GWM, Fernandez-Vidal S, Karachi C, Bonneville F, Do MC, Delmaire C, Dormont D, Bardinet E, Agid Y, Chastan N, Welter ML (2014). High-level gait and balance disorders in the elderly: a midbrain disease? *J Neurol* 261:196–206. <https://doi.org/10.1007/s00415-013-7174-x>
- Diakhaté DG, Do MC, Le Bozec S (2013). Effects of seat-thigh contact on kinematics performance in sit-to-stand and trunk flexion tasks. *J Biomech* 46:879–882. <https://doi.org/10.1016/j.jbiomech.2012.12.022>
- Dick JP, Rothwell JC, Berardelli A, Thompson PD, Gioux M, Benecke R, Day BL, Marsden CD (1986). Associated postural adjustments in Parkinson's disease. *J Neurol Neurosurg Psychiatry* 49:1378–1385
- Do MC, Nouillot P, Bouisset S (1991). Is balance or posture at the end of a voluntary movement programmed? *Neuroscience letters* 130:9–11. [https://doi.org/10.1016/0304-3940\(91\)90215-F](https://doi.org/10.1016/0304-3940(91)90215-F)
- Donelan JM, Shipman D, Kram R, Kuo AD (2004). Mechanical and metabolic requirements for active lateral stabilization in human walking. *J Biomech* 37:827–835. <https://doi.org/10.1016/j.jbiomech.2003.06.002>
- Dorfman LJ, Bosley TM (1979). Age-related changes in peripheral and central nerve conduction in man. *Neurology* 29(1):38-44. doi: 10.1212/wnl.29.1.38
- Elftman H (1939). The function of arms in walking. *Hum Biol* 11(4):529—35.
- Fischer M, Vialleron T, Laffaye G, Fourcade P, Hussein T, Chèze L, Deleu PA, Honeine JL, Yiou E, Delafontaine A (2019). Long-Term Effects of Whole-Body Vibration on Human Gait: A Systematic Review and Meta-Analysis. *Front Neurol* 10:627. <https://doi.org/10.3389/fneur.2019.00627>
- Floyd WF (1955). Silver PHS. The function of Erectores Spinae muscles in certain movements and postures in man. *J Physiol Lond* 129(1):184—203.
- Fourcade P, Bouisset S, Le Bozec S, Memari S (2018). Consecutive postural adjustments (CPAs): A kinetic analysis of variable velocity during a pointing task. *Neurophysiologie Clinique* 48:387–396. <https://doi.org/10.1016/j.neucli.2018.01.004>
- Fourcade P, Hansen C, LeBozec S, Bouisset S (2014). Simultaneous Postural Adjustments (SPA) scrutinized using the Lissajous method. *J Biomech* 47:3645–3649. <https://doi.org/10.1016/j.jbiomech.2014.10.004>

- Fourcade P, Le Bozec S, Bouisset S (2016). Are simultaneous postural adjustments (SPA) programmed as a function of pointing velocity? *Exp Brain Res* 234:2809–2817. <https://doi.org/10.1007/s00221-016-4683-8>
- Fourcade P, Memari S (2015). Simultaneous postural adjustments (SPA) in single step, analysed using the “Lissajous method.” *Comput Methods Biomech Biomed Engin* 18 Suppl 1:1938–1939. <https://doi.org/10.1080/10255842.2015.1069577>
- Fowler BD, Palombo KTM, Feland JB, Blotter JD (2019). Effects of Whole-Body Vibration on Flexibility and Stiffness: A Literature Review. *Int J Exerc Sci* 12:735–747
- Frazzitta G, Maestri R, Bertotti G, Riboldazzi G, Boveri N, Perini M, Uccellini D, Turla M, Comi C, Pezzoli G, Ghilardi MF (2015). Intensive rehabilitation treatment in early Parkinson’s disease: a randomized pilot study with a 2-year follow-up. *Neurorehabil Neural Repair* 29:123–131. <https://doi.org/10.1177/1545968314542981>
- Friedli WG, Cohen L, Hallett M, Stanhope S, Simon SR (1988). Postural adjustments associated with rapid voluntary arm movements. II. Biomechanical analysis. *Journal of Neurology, Neurosurgery, and Psychiatry* 51:232. <https://doi.org/10.1136/jnnp.51.2.232>
- Gahéry Y. (1987). Associated movements, postural adjustments and synergies: some comments about the history and significance of three motor concepts. *Arch Ital Biol.*125:345-60.
- Gajdosik RL, Vander Linden DW, McNair PJ, Williams AK, Riggin TJ (2005). Effects of an eight-week stretching program on the passive-elastic properties and function of the calf muscles of older women. *Clinical Biomechanics* 20:973–983. <https://doi.org/10.1016/j.clinbiomech.2005.05.011>
- Gatev P, Thomas S, Kepple T, Hallett M (1999). Feedforward ankle strategy of balance during quiet stance in adults. *J Physiol* 514:915–928. <https://doi.org/10.1111/j.1469-7793.1999.915ad.x>
- Gendre M, Yiou E, Gélat T, Honeine JL, Deroche T (2016). Directional specificity of postural threat on anticipatory postural adjustments during lateral leg raising. *Exp Brain Res* 234:659–671. <https://doi.org/10.1007/s00221-015-4471-x>
- Goutal L, Lino F, Bouisset S (1994). Modalités de l’appui corporel et vitesse du mouvement de pointage. *Arch Int Physiol Bioch Biophys.*102:C21.
- Gray J. *Animal locomotion* (1968). Weidenfeld and Nicolson, London.
- Gurfinkel VS, Fomin SV, Tsetlin ML (1971). The compensation of respiratory disturbances of the erect posture of man as an example of the organization of

- interarticular interaction. *Models of the Structural Functional Organization of Certain Biological Systems*. MIT Press, Cambridge: 382-395
- Hagbarth KE, Eklund G (1966). Tonic vibration reflexes (TVR) in spasticity. *Brain Res* 2:201–203
- Hahn ME, Chou LS (2004). Age-related reduction in sagittal plane center of mass motion during obstacle crossing. *J Biomech* 37:837–844. <https://doi.org/10.1016/j.jbiomech.2003.11.010>
- Halliday SE, Winter DA, Frank JS, Patla AE, Prince F (1998). The initiation of gait in young, elderly, and Parkinson's disease subjects. *Gait Posture* 8:8–14. [https://doi.org/10.1016/s0966-6362\(98\)00020-4](https://doi.org/10.1016/s0966-6362(98)00020-4)
- Hamaoui A, Alamini-Rodrigues C (2017). Influence of Cervical Spine Mobility on the Focal and Postural Components of the Sit-to-Stand Task. *Front Hum Neurosci* 11:129. <https://doi.org/10.3389/fnhum.2017.00129>
- Hamaoui A, Do M c, Poupard L, Bouisset S (2002). Does respiration perturb body balance more in chronic low back pain subjects than in healthy subjects? *Clin Biomech (Bristol, Avon)* 17:548–550. [https://doi.org/10.1016/s0268-0033\(02\)00042-6](https://doi.org/10.1016/s0268-0033(02)00042-6)
- Hamaoui A, Do MC, Bouisset S (2004). Postural sway increase in low back pain subjects is not related to reduced spine range of motion. *Neurosci Lett* 357:135–138. <https://doi.org/10.1016/j.neulet.2003.12.047>
- Han SW, Lee DY, Choi DS, Han B, Kim JS, Lee HD (2017). Asynchronous Alterations of Muscle Force and Tendon Stiffness Following 8 Weeks of Resistance Exercise with Whole-Body
- Hansen C, Wei Q, Shieh JS, Fourcade P, Isableu B, Majed L. (2017). Sample Entropy, Univariate, and Multivariate Multi-Scale Entropy in Comparison with Classical Postural Sway Parameters in Young Healthy Adults. *Front Hum Neurosci*.11:206. <https://doi.org/10.1123/japa.2016-0149>
- Hass CJ, Buckley TA, Pitsikoulis C, Barthelemy EJ (2012). Progressive resistance training improves gait initiation in individuals with Parkinson's disease. *Gait Posture* 35:669–673. <https://doi.org/10.1016/j.gaitpost.2011.12.022>
- Hass CJ, Gregor RJ, Waddell DE, Oliver A, Smith DW, Fleming RP, Wolf SL (2004). The influence of Tai Chi training on the center of pressure trajectory during gait initiation in older adults. *Arch Phys Med Rehabil* 85:1593–1598. <https://doi.org/10.1016/j.apmr.2004.01.020>

- Hausdorff JM, Balash Y, Giladi N (2003). Time series analysis of leg movements during freezing of gait in Parkinson's disease: akinesia, rhyme or reason? *Physica A: Statistical Mechanics and its Applications* 321:565–570. [https://doi.org/10.1016/S0378-4371\(02\)01744-2](https://doi.org/10.1016/S0378-4371(02)01744-2)
- Henriksson M, Henriksson J, Bergenius J (2011). Gait initiation characteristics in elderly patients with unilateral vestibular impairment. *Gait & Posture* 33:661–667. <https://doi.org/10.1016/j.gaitpost.2011.02.018>
- Hess WR. (1943). Teleokinetisches und ereismatisches Kräftesystem in der Biomotorik. *Helv Physiol Pharmacol Acta*.1:C62-C63.
- Hof AL, Gazendam MGJ, Sinke WE (2005). The condition for dynamic stability. *J Biomech* 38:1–8. <https://doi.org/10.1016/j.jbiomech.2004.03.025>
- Hof AL, van Bockel RM, Schoppen T, Postema K (2007). Control of lateral balance in walking. Experimental findings in normal subjects and above-knee amputees. *Gait Posture* 25:250–258. <https://doi.org/10.1016/j.gaitpost.2006.04.013>
- Honeine JL, Schieppati M, Crisafulli O, Do MC (2016). The Neuro-Mechanical Processes That Underlie Goal-Directed Medio-Lateral APA during Gait Initiation. *Front Hum Neurosci* 10: 445. <https://doi.org/10.3389/fnhum.2016.00445>
- Honeine JL, Schieppati M, Gagey O, Do MC (2014). By counteracting gravity, triceps surae sets both kinematics and kinetics of gait. *Physiol Rep* 2:e00229. <https://doi.org/10.1002/phy2.229>
- Honeine JL, Schieppati M, Gagey O, Do MC (2013). The Functional Role of the Triceps Surae Muscle during Human Locomotion. *PLoS One* 8: e52943. <https://doi.org/10.1371/journal.pone.0052943>
- Horak FB, Esselman P, Anderson ME, Lynch MK (1984). The effects of movement velocity, mass displaced, and task certainty on associated postural adjustments made by normal and hemiplegic individuals. *J Neurol Neurosurg Psychiatry* 4:1020– 1028. doi: 10.1136/jnnp.47.9.1020
- Horak FB, Nashner LM (1986). Central programming of postural movements: adaptation to altered support-surface configurations. *J Neurophysiol* 55(6):1369-81 <https://doi.org/10.1152/jn.1986.55.6.1369>
- Horak FB, Shupert CL, Mirka A (1989). Components of postural dyscontrol in the elderly: a review. *Neurobiol Aging* 10(6):727-38. doi: 10.1016/0197-4580(89)90010-9
- Hotta K, Batchelor WB, Graven J, Dahya V, Noel TE, Ghai A, Katopodis JN, Dixon WC, Andrews R, Pragle A, Chheda J, Liberatore L, Behnke BJ, Muller-Delp J (2019). Daily Passive Muscle Stretching Improves Flow-Mediated Dilation of Popliteal

- Artery and 6-minute Walk Test in Elderly Patients with Stable Symptomatic Peripheral Artery Disease. *Cardiovasc Revasc Med* 20:642–648. <https://doi.org/10.1016/j.carrev.2019.05.003>
- Ilmane N, LaRue J (2008). Modulation of anticipatory postural adjustments in a complex task under different temporal constraints. *Motor Control* 12(4):330-47. doi: 10.1123/mcj.12.4.330.
- Ilmane N, LaRue J (2011). Modulation of Anticipatory Postural Adjustments in the Anticipation-Coincidence Task. *Journal of Motor Behavior* 43:333–343. <https://doi.org/10.1080/00222895.2011.594819>
- Jacobs JV, Lou JS, Kraakevik JA, Horak FB (2009a). The supplementary motor area contributes to the timing of the anticipatory postural adjustment during step initiation in participants with and without Parkinson's disease. *Neuroscience* 164:877–885. <https://doi.org/10.1016/j.neuroscience.2009.08.002>
- Jacobs JV, Nutt JG, Carlson-Kuhta P, Stephens M, Horak FB (2009b). Knee trembling during freezing of gait represents multiple anticipatory postural adjustments. *Exp Neurol* 215:334–341. <https://doi.org/10.1016/j.expneurol.2008.10.019>
- Jian Y, Winter D, Ishaq M, Gilchrist L (1993). Trajectory of the body COG and COP during initiation and termination of gait. *Gait & Posture* 1:9–22. [https://doi.org/10.1016/0966-6362\(93\)90038-3](https://doi.org/10.1016/0966-6362(93)90038-3)
- Kanekar N, Aruin AS (2014). The effect of aging on anticipatory postural control. *Exp Brain Res* 232:1127–1136. <https://doi.org/10.1007/s00221-014-3822-3>
- Kanekar N, Aruin AS (2015). Improvement of anticipatory postural adjustments for balance control: effect of a single training session. *J Electromyogr Kinesiol* 25:400–405. <https://doi.org/10.1016/j.jelekin.2014.11.002>
- Kararizou E, Manta P, Kalfakis N, Vassilopoulos D (2005). Morphometric study of the human muscle spindle. *Anal Quant Cytol Histol* 27(1):1-4.
- Khanmohammadi R, Talebian S, Hadian MR, Olyaei G, Bagheri H (2015). Preparatory postural adjustments during gait initiation in healthy younger and older adults: Neurophysiological and biomechanical aspects. *Brain Res* 1629:240–249. <https://doi.org/10.1016/j.brainres.2015.09.039>
- King DL, Zatsiorsky VM (2002). Periods of extreme ankle displacement during one-legged standing. *Gait Posture* 15:172–179
- Krishnan V, Vennila K, Aruin AS (2011). Postural control in response to a perturbation: role of vision and additional support. *Exp Brain Res* 212:385–397. <https://doi.org/10.1007/s00221-011-2738-4>



- Kuo AD (1995). An optimal control model for analyzing human postural balance. *IEEE Trans Biomed Eng* 42:87–101. <https://doi.org/10.1109/10.362914>
- Kuo AD (2007). The six determinants of gait and the inverted pendulum analogy: A dynamic walking perspective. *Hum Mov Sci* 26:617–656. <https://doi.org/10.1016/j.humov.2007.04.003>
- Kurtzer I, Herter TM, Scott SH (2005). Random change in cortical load representation suggests distinct control of posture and movement. *Nat Neurosci.* 8(4):498–504. doi: 10.1038/nn1420. Epub 2005 Mar 13
- Lam FM, Chan PF, Liao L, Woo J, Hui E, Lai CWK, Kwok TC, Pang MYC (2018). Effects of whole-body vibration on balance and mobility in institutionalized older adults: a randomized controlled trial. *Clinical Rehabilitation* 32:462–472. <https://doi.org/10.1177/0269215517733525>
- Le Bozec S, Bouisset S (2004). Does postural chain mobility influence muscular control in sitting ramp pushes? *Exp Brain Res* 158:427–437. <https://doi.org/10.1007/s00221-004-1918-x>
- Le Bozec S, Bouisset S, Ribreau C (2008). Postural control in isometric ramp pushes: the role of Consecutive Postural Adjustments (CPAs). *Neurosci Lett* 448:250–254. <https://doi.org/10.1016/j.neulet.2008.10.047>
- Lee A, Gilbert RM (2016). Epidemiology of Parkinson Disease. *Neurologic Clinics* 34:955–965. <https://doi.org/10.1016/j.ncl.2016.06.012>
- Lee PG, Jackson EA, Richardson CR (2017). Exercise Prescriptions in Older Adults. 95:8
- Lee WA, Buchanan TS, Rogers MW (1987). Effects of arm acceleration and behavioral conditions on the organization of postural adjustments during arm flexion. *Exp Brain Res* 66:257–270. <https://doi.org/10.1007/BF00243303>
- Lepers R, Brenière Y (1995). The role of anticipatory postural adjustments and gravity in gait initiation. *Exp Brain Res* 107:118–124. <https://doi.org/10.1007/bf00228023>
- Li L, Lamis F, Wilson SE (2008). Whole-body vibration alters proprioception in the trunk. *International Journal of Industrial Ergonomics* 38:792–800. <https://doi.org/10.1016/j.ergon.2007.10.010>
- Lieberman DE, Venkadesan M, Werbel WA, Daoud AI, D'Andrea S, Davis IS, Mang'Eni RO, Pitsiladis Y (2010). Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature* 463:531–535. <https://doi.org/10.1038/nature08723>

- Lin SI, Yang WC (2011). Effect of plantar desensitization on postural adjustments prior to step initiation. *Gait Posture* 34:451–456. <https://doi.org/10.1016/j.gaitpost.2011.06.016>
- Lino F, Duchêne FL, Bouisset S (1992). Effect of seat contact area on the velocity of a pointing task, In: *Biomechanics*. P. Belloti, and A. Capozzo (Ed.) Università La Sapienza; pp. 232.
- Lu C, Amundsen Huffmaster SL, Harvey JC, MacKinnon CD (2017). Anticipatory postural adjustment patterns during gait initiation across the adult lifespan. *Gait Posture* 57:182–187. <https://doi.org/10.1016/j.gaitpost.2017.06.010>
- Luccarini P, Gahery Y, Pompeiano O (1990). Cholinoceptive pontine reticular structures modify the postural adjustments during the limb movements induced by cortical stimulation. *Arch Ital Biol* 128:19–45
- Lundervold A (1951). Electromyographic investigations of position and manner of working in typewriting. *Acta Physiol Scand suppl* 84:1–171.
- Lyon IN, Day BL (1997). Control of frontal plane body motion in human stepping. *Exp Brain Res* 115:345–356
- Lyon IN, Day BL (2005). Predictive control of body mass trajectory in a two-step sequence. *Experimental Brain Research* 161:193–200. <https://doi.org/10.1007/s00221-004-2058-z>
- MacKinnon CD, Winter DA (1993). Control of whole body balance in the frontal plane during human walking. *J Biomech* 26:633–644. [https://doi.org/10.1016/0021-9290\(93\)90027-c](https://doi.org/10.1016/0021-9290(93)90027-c)
- Madhavan S, Shields RK (2005). Influence of age on dynamic position sense: evidence using a sequential movement task. *Exp. Brain Res.* 164(1):18–28. doi: 10.1007/s00221-004-2208-3.
- Maki BE, Edmondstone MA, McIlroy WE (2000). Age-related differences in laterally directed compensatory stepping behavior. *J Gerontol A Biol Sci Med Sci* 55:M270–277. <https://doi.org/10.1093/gerona/55.5.m270>
- Maki BE, McIlroy WE (2006). Control of rapid limb movements for balance recovery: age-related changes and implications for fall prevention. *Age Ageing* 35 Suppl 2:ii12–ii18. <https://doi.org/10.1093/ageing/af1078>
- Mancini M, Zampieri C, Carlson-Kuhta P, Chiari L, Horak FB (2009). Anticipatory postural adjustments prior to step initiation are hypometric in untreated parkinson's disease: an accelerometer-based approach. *Eur J Neurol* 16:1028–1034. <https://doi.org/10.1111/j.1468-1331.2009.02641.x>

- Mann RA, Hagy JL, White V, Liddell D (1979). The initiation of gait. *J Bone Joint Surg Am* 61:232–239
- Marigold DS (2008). Role of peripheral visual cues in online visual guidance of locomotion. *Exercise and Sport Sciences Reviews* 36:145–151. <https://doi.org/10.1097/JES.0b013e31817bff72>
- Marinho-Buzelli AR, Barela AMF, Barela JA, Celestino ML, Popovic MR, Verrier M (2017a). The Influence of the Aquatic Environment on Gait Initiation: A Pilot Study. *Motor Control* 21:211–226. <https://doi.org/10.1123/mc.2015-0091>
- Marinho-Buzelli AR, Masani K, Rouhani H, Barela AM, Fernandes GTB, Verrier MC, Popovic MR (2017b). The influence of the aquatic environment on the center of pressure, impulses and upper and lower trunk accelerations during gait initiation. *Gait Posture* 58:469–475. <https://doi.org/10.1016/j.gaitpost.2017.09.008>
- Martínez-Velilla N, Cadore EL, Casas-Herrero Á, Idoate-Saralegui F, Izquierdo M (2016). Physical Activity and Early Rehabilitation in Hospitalized Elderly Medical Patients: Systematic Review of Randomized Clinical Trials. *J Nutr Health Aging* 20:738–751. <https://doi.org/10.1007/s12603-016-0683-4>
- Massion J (1992). Movement, posture and equilibrium: Interaction and coordination. *Progress in Neurobiology* 38:35–56. [https://doi.org/10.1016/0301-0082\(92\)90034-C](https://doi.org/10.1016/0301-0082(92)90034-C)
- Massion J (1994). Postural control system. *Curr Opin Neurobiol* 4(6):877–87. doi: 10.1016/0959-4388(94)90137-6
- Massion J, Alexandrov A, Frolov A, (2004). Why and how are posture and movement coordinated? *Prog. Brain Res.* 143: 13–27. [https://doi.org/10.1016/S0079-6123\(03\)43002-1](https://doi.org/10.1016/S0079-6123(03)43002-1)
- McIlroy WE, Maki BE (1999). The control of lateral stability during rapid stepping reactions evoked by antero-posterior perturbation: does anticipatory control play a role? *Gait Posture* 9:190–198
- Memari S, Do MC, Le Bozec S, Bouisset S (2013). The Consecutive Postural Adjustments (CPAs) that follow foot placement in single stepping. *Neurosci Lett* 543:32–36. <https://doi.org/10.1016/j.neulet.2013.02.061>
- Memari S, Yiou E, Fourcade P (2020). The role(s) of "Simultaneous Postural Adjustments" (SPA) during Single Step revealed with the Lissajous method. *J Biomech* 108:109910. doi:10.1016/j.jbiomech.2020.109910

- Mora JC, Valencia WM (2018). Exercise and Older Adults. *Clinics in Geriatric Medicine* 34:145–162. <https://doi.org/10.1016/j.cger.2017.08.007>
- Morisod J, Coutaz M (2007). Post-fall syndrome: how to recognize and treat it?
- Mouchnino L, Blouin J (2013). When Standing on a Moving Support, Cutaneous Inputs Provide Sufficient Information to Plan the Anticipatory Postural Adjustments for Gait Initiation. *PLoS One* 8:e55081. <https://doi.org/10.1371/journal.pone.0055081>
- Mouchnino L, Robert G, Ruget H, Blouin J, Simoneau M (2012). Online control of anticipated postural adjustments in step initiation: Evidence from behavioral and computational approaches. *Gait & Posture* 35:616–620. <https://doi.org/10.1016/j.gaitpost.2011.12.009>
- Muir BC, Rietdyk S, Haddad JM (2014). Gait initiation: the first four steps in adults aged 20-25 years, 65-79 years, and 80-91 years. *Gait Posture* 39:490–494. <https://doi.org/10.1016/j.gaitpost.2013.08.037>
- Muyinat YO, Ashwini KR, Sunil KA, Anil KL (2019). Balance and gait in the elderly: A contemporary review. *Laryngoscope Investig Otolaryngol* 4(1):143-153. doi:10.1002/lio2.252. eCollection 2019 Feb.
- Nardone A, Schieppati M (1988). Postural adjustments associated with voluntary contraction of leg muscles in standing man. *Exp Brain Res* 69:469–480. <https://doi.org/10.1007/bf00247301>
- Ninot G (2019). Guide professionnel des interventions non médicamenteuses
- Nougier V, Teasdale N, Bard C, Fleury M (1999). Modulation of anticipatory postural adjustments in a reactive and a self-triggered mode in humans. *Neurosci Lett* 260:109–112. [https://doi.org/10.1016/s0304-3940\(98\)00961-6](https://doi.org/10.1016/s0304-3940(98)00961-6)
- Nouillot P, Bouisset S, Do MC (1992). Do fast voluntary movements necessitate anticipatory postural adjustments even if equilibrium is unstable? *Neurosci Lett* 147:1–4
- O'Connor SM, Kuo AD (2009). Direction-dependent control of balance during walking and standing. *J Neurophysiol* 102:1411–1419. <https://doi.org/10.1152/jn.00131.2009>
- Oddsson L, Thorstensson A (1987). Fast voluntary trunk flexion movements in standing: motor patterns. *Acta Physiol Scand* 129:93–106. <https://doi.org/10.1111/j.1748-1716.1987.tb08044.x>

- Okada Y, Fukumoto T, Takatori K, Nagino K, Hiraoka K (2011a). Variable initial swing side and prolonged double limb support represent abnormalities of the first three steps of gait initiation in patients with Parkinson's disease with freezing of gait. *Front Neurol* 2:85. <https://doi.org/10.3389/fneur.2011.00085>
- Okada Y, Fukumoto T, Takatori K, Nagino K, Hiraoka K (2011b). Abnormalities of the First Three Steps of Gait Initiation in Patients with Parkinson's Disease with Freezing of Gait. *Parkinson's disease* 2011:202937. <https://doi.org/10.4061/2011/202937>
- Onambele GL, Narici MV, Maganaris CN (2006). Calf muscle-tendon properties and postural balance in old age. *J Appl Physiol* (1985) 100(6):2048-56. doi:10.1152/jappphysiol.01442.2005
- Orr R (2015). The effect of whole body vibration exposure on balance and functional mobility in older adults: a systematic review and meta-analysis. *Maturitas* 80:342–358. <https://doi.org/10.1016/j.maturitas.2014.12.020>
- Pai YC, Patton J (1997). Center of mass velocity-position predictions for balance control. *J Biomech* 30:347–354. [https://doi.org/10.1016/s0021-9290\(96\)00165-0](https://doi.org/10.1016/s0021-9290(96)00165-0)
- Paillard J. (1990). Réactif et prédictif : Deux modes de gestion de la motricité. *Rev. Neurol. (Paris)* 146: 600–611.
- Park SY, Son WM, Kwon OS (2015). Effects of whole body vibration training on body composition, skeletal muscle strength, and cardiovascular health. *J Exerc Rehabil* 11:289–295. <https://doi.org/10.12965/jer.150254>
- Patchay S, Gahéry Y, Serratrice G (1997). Gait initiation and impairments of ground reaction forces as illustrated in old age by 'La marche à petits pas'. *Neurosci Lett* 236(3):143-6. doi: 10.1016/s0304-3940(97)00781-7.
- Patchay S, Gahéry Y, Serratrice G (2002). Early postural adjustments associated with gait initiation and age-related walking difficulties. *Mov Disord* 17:317–326
- Pedotti A, Crenna P, Deat A, Frigo C, Massion J (1989). Postural synergies in axial movements: short and long-term adaptation. *Exp Brain Res* 74:3–10. <https://doi.org/10.1007/bf00248275>
- Perera S, Patel KV, Rosano C, Rubin SM, Satterfield S, Harris T, Ensrud K, Orwoll E, Lee CG, Chandler JM, Newman AB, Cauley JA, Guralnik JM, Ferrucci L, Studenski SA (2016). Gait Speed Predicts Incident Disability: A Pooled Analysis. *J Gerontol A Biol Sci Med Sci* 71:63–71. <https://doi.org/10.1093/gerona/glv126>
- Peters RM, McKeown MD, Carpenter MG, Inglis JT (2016). Losing touch: age-related changes in plantar skin sensitivity, lower limb cutaneous reflex strength, and

- postural stability in older adults. *J Neurophysiol* 116:1848–1858. <https://doi.org/10.1152/jn.00339.2016>
- Pirker W, Katzenschlager R (2017). Gait disorders in adults and the elderly: A clinical guide. *Wien Klin Wochenschr* 129:81–95. <https://doi.org/10.1007/s00508-016-1096-4>
- Pollock RD, Provan S, Martin FC, Newham DJ (2010). Whole body vibration; short-term effects on balance, proprioception and cutaneous sensation in healthy young adults. *Proc of The Phys Soc Proc Physiol Soc* 19:
- Pollock RD, Provan S, Martin FC, Newham DJ (2011). The effects of whole body vibration on balance, joint position sense and cutaneous sensation. *Eur J Appl Physiol* 111:3069–3077. <https://doi.org/10.1007/s00421-011-1943-y>
- Portero P, McNair P (2015). Les étirements musculo-tendineux: des données scientifiques à une pratique raisonnée. *Kinésithérapie, la Revue* 15:32–40. <https://doi.org/10.1016/j.kine.2015.06.009>
- Raza C, Anjum R, Ain Shakeel NU (2019). Parkinson's disease: Mechanisms, translational models and management strategies *Life Sciences* 226:77-90. doi: 10.1016/j.lfs.2019.03.057
- Radford JA, Burns J, Buchbinder R, Landorf KB, Cook C (2006). Does stretching increase ankle dorsiflexion range of motion? A systematic review. *Br J Sports Med* 40:870–875. <https://doi.org/10.1136/bjism.2006.029348>
- Rankin BL, Buffo SK, Dean JC (2014). A neuromechanical strategy for mediolateral foot placement in walking humans. *J Neurophysiol* 112:374–383. <https://doi.org/10.1152/jn.00138.2014>
- Reimann H, Fettrow TD, Thompson ED, Agada P, McFadyen BJ, Jeka JJ (2017). Complementary mechanisms for upright balance during walking. *PLoS ONE* 12:e0172215. <https://doi.org/10.1371/journal.pone.0172215>
- Reynolds RF, Day BL (2005). Visual guidance of the human foot during a step. *The Journal of Physiology* 569:677–684. <https://doi.org/10.1113/jphysiol.2005.095869>
- Richard A, Van Hamme A, Drevelle X, Golmard JL, Meunier S, Welter ML (2017). Contribution of the supplementary motor area and the cerebellum to the anticipatory postural adjustments and execution phases of human gait initiation. *Neuroscience* 358:181–189. <https://doi.org/10.1016/j.neuroscience.2017.06.047>

- Ritzmann R, Gollhofer A, Kramer A (2013). The influence of vibration type, frequency, body position and additional load on the neuromuscular activity during whole body vibration. *Eur J Appl Physiol.* 113:1–11. doi: 10.1007/s00421-012-2402-0
- Robert G, Blouin J, Ruget H, Mouchnino L (2007). Coordination between postural and movement controls: effect of changes in body mass distribution on postural and focal component characteristics. *Exp Brain Res* 181:159–171. <https://doi.org/10.1007/s00221-007-0916-1>
- Robinovitch SN, Feldman F, Yang Y, Schonnop R, Leung PM, Sarraf T, Sims-Gould J, Loughin M (2013). Video capture of the circumstances of falls in elderly people residing in long-term care: an observational study. *Lancet* 381:47–54. [https://doi.org/10.1016/S0140-6736\(12\)61263-X](https://doi.org/10.1016/S0140-6736(12)61263-X)
- Robinovitch SN, Inkster L, Maurer J, Warnick B (2003). Strategies for avoiding hip impact during sideways falls. *J Bone Miner Res* 18:1267–1273. <https://doi.org/10.1359/jbmr.2003.18.7.1267>
- Rocchi L, Carlson-Kuhta P, Chiari L, Burchiel KJ, Hogarth P, Horak FB (2012). Effects of deep brain stimulation in the subthalamic nucleus or globus pallidus internus on step initiation in Parkinson disease: laboratory investigation. *J Neurosurg* 117:1141–1149. <https://doi.org/10.3171/2012.8.JNS112006>
- Rochester L, Galna B, Lord S, Burn D (2014). The nature of dual-task interference during gait in incident Parkinson's disease. *Neuroscience* 265:83–94. <https://doi.org/10.1016/j.neuroscience.2014.01.041>
- Roden-Reynolds DC, Walker MH, Wasserman CR, Dean JC (2015). Hip proprioceptive feedback influences the control of mediolateral stability during human walking. *J Neurophysiol* 114:2220–2229. <https://doi.org/10.1152/jn.00551.2015>
- Rogan S, Taeymans J, Radlinger L, Naepflin S, Ruppen S, Bruehlhart Y, Hilfiker R (2017). Effects of whole-body vibration on postural control in elderly: An update of a systematic review and meta-analysis. *Arch Gerontol Geriatr* 73:95–112. <https://doi.org/10.1016/j.archger.2017.07.022>
- Rogers MW, Pai YC (1990). Dynamic transitions in stance support accompanying leg flexion movements in man. *Exp Brain Res* 81:398–402. <https://doi.org/10.1007/bf00228132>
- Romberg MH. *Manuel of nervous disease in man*(1953). London, Sydeham Society,:395-401.
- Roschel H, Barroso R, Tricoli V, Batista MAB, Acquesta FM, Serrão JC, Ugrinowitsch C (2015). Effects of Strength Training Associated With Whole-Body Vibration

- Training on Running Economy and Vertical Stiffness. *J Strength Cond Res* 29:2215–2220. <https://doi.org/10.1519/JSC.0000000000000857>
- Ruget H, Blouin J, Coyle T, Mouchnino L (2010). Modulation of proprioceptive inflow when initiating a step influences postural adjustments. *Exp Brain Res* 201:297–305. <https://doi.org/10.1007/s00221-009-2035-7>
- Ruget H, Blouin J, Teasdale N, Mouchnino L (2008). Can prepared anticipatory postural adjustments be updated by proprioception? *Neuroscience* 155:640–648. <https://doi.org/10.1016/j.neuroscience.2008.06.021>
- Santos MJ, Kanekar N, Aruin AS (2010). The Role of Anticipatory Postural Adjustments in Compensatory Control of Posture: 2. Biomechanical Analysis. *J Electromyogr Kinesiol* 20:398–405. <https://doi.org/10.1016/j.jelekin.2010.01.002>
- Saradjian AH, Tremblay L, Perrier J, Blouin J, Mouchnino L (2013). Cortical facilitation of proprioceptive inputs related to gravitational balance constraints during step preparation. *J Neurophysiol* 110:397–407. <https://doi.org/10.1152/jn.00905.2012>
- Sasaki O, Asawa S, Katsuno S, Usami S, Taguchi K (2001). Gait initiation in bilateral vestibular loss. *Auris Nasus Larynx* 28:295–299
- Schaafsma JD, Balash Y, Gurevich T, Bartels AL, Hausdorff JM, Giladi N (2003). Characterization of freezing of gait subtypes and the response of each to levodopa in Parkinson's disease. *European Journal of Neurology* 10:391–398. <https://doi.org/10.1046/j.1468-1331.2003.00611.x>
- Shaffer SW, Harrison AL (2007). Aging of the Somatosensory System: A Translational Perspective. *Phys. Ther.* 87, 193–207. doi: 10.2522/ptj.20060083
- Schepens B, Drew T (2006). Descending Signals From the Pontomedullary Reticular Formation Are Bilateral, Asymmetric, and Gated During Reaching Movements in the Cat. *Journal of Neurophysiology* 96:2229–2252. <https://doi.org/10.1152/jn.00342.2006>
- Schepens B, Drew T (2003). Strategies for the Integration of Posture and Movement During Reaching in the Cat. *Journal of Neurophysiology* 90:3066–3086. <https://doi.org/10.1152/jn.00339.2003>
- Schieppati M, Hugon M, Grasso M, Nardone A, Galante M (1994). The limits of equilibrium in young and elderly normal subjects and in parkinsonians. *Electroencephalogr Clin Neurophysiol* 93:286–298. [https://doi.org/10.1016/0168-5597\(94\)90031-0](https://doi.org/10.1016/0168-5597(94)90031-0)



- Schlee G, Reckmann D, Milani TL (2012). Whole body vibration training reduces plantar foot sensitivity but improves balance control of healthy subjects. *Neurosci Lett* 506:70–73. <https://doi.org/10.1016/j.neulet.2011.10.051>
- Schniepp R, Wuehr M, Pradhan C, Novozhilov S, Krafczyk S, Brandt T, Jahn K (2013). Nonlinear variability of body sway in patients with phobic postural vertigo. *Front Neurol* 4:115. <https://doi.org/10.3389/fneur.2013.00115>
- Simuni T, Stern MB (1999). Does levodopa accelerate Parkinson's disease? *Drugs Aging* 14(6):399–408. doi: 10.2165/00002512-199914060-00001
- Skinner HB, Barrack RL, Cook SD (1984). Age-related decline in proprioception. *Clin Orthop Relat Res* (184):208–11.
- Slijper H, Latash ML, Mordkoff JT (2002). Anticipatory postural adjustments under simple and choice reaction time conditions. *Brain Res* 924:184–197. [https://doi.org/10.1016/s0006-8993\(01\)03233-4](https://doi.org/10.1016/s0006-8993(01)03233-4)
- Smith DT, Judge S, Malone A, Moynes RC, Conviser J, Skinner JS (2016). Effects of bioDensity Training and Power Plate Whole-Body Vibration on Strength, Balance, and Functional Independence in Older Adults. *J Aging Phys Act* 24:139–148. <https://doi.org/10.1123/japa.2015-0057>
- Stapley P, Pozzo T, Chéron G, Grishin A (1999). Does the coordination between posture and movement during whole body reaching task ensure centre of mass stabilization? *Exp Brain Res* 129:134–146. <https://doi.org/10.1007/s002210050944>
- Stapley P, Pozzo T, Grishin A (1998). The role of anticipatory adjustments during whole body reaching movements. *Neuroreport* 9:395–401. doi: 10.1097/00001756-199802160-00007.
- Tambasco N, Romoli M, Calabresi P (2018). Levodopa in Parkinson's Disease: Current Status and Future Developments. *Curr Neuropharmacol* 16(8):1239–1252. doi: 10.2174/1570159X15666170510143821
- Takakusaki K (2017). Functional Neuroanatomy for Posture and Gait Control. *J Mov Disord* 10(1):1–17. doi: 10.14802/jmd.16062
- Teyssèdre C, Lino F, Zattara M, Bouisset S (2000). Anticipatory EMG patterns associated with preferred and non-preferred arm pointing movements. *Exp Brain Res* 134:435–440. <https://doi.org/10.1007/s002210000490>
- Thoumie P, Do MC (1996). Changes in motor activity and biomechanics during balance recovery following cutaneous and muscular deafferentation. *Exp Brain Res* 110:289–297

- Townsend MA (1985). Biped gait stabilization via foot placement. *J Biomech* 18:21–38. [https://doi.org/10.1016/0021-9290\(85\)90042-9](https://doi.org/10.1016/0021-9290(85)90042-9)
- Tysnes OB, Storstein A (2017). Epidemiology of Parkinson's disease. *Journal of Neural Transmission* 124:901–905. <https://doi.org/10.1007/s00702-017-1686-y>
- Uemura K, Yamada M, Nagai K, Tanaka B, Mori S, Ichihashi S (2012a). Fear of falling is associated with prolonged anticipatory postural adjustment during gait initiation under dual-task conditions in older adults. *Gait Posture* 35(2):282–6. doi: 10.1016/j.gaitpost.2011.09.100.
- Uemura K, Yamada M, Nagai K, Tateuchi H, Mori S, Tanaka B, Ichihashi N (2012b). Effects of dual-task switch exercise on gait and gait initiation performance in older adults: preliminary results of a randomized controlled trial. *Arch Gerontol Geriatr* 54:e167–171. <https://doi.org/10.1016/j.archger.2012.01.002>
- Uemura K, Yamada M, Nagai K, Ichihashi N (2011). Older adults at high risk of falling need more time for anticipatory postural adjustment in the precrossing phase of obstacle negotiation. *J Gerontol A Biol Sci Med Sci* 66:904–909. <https://doi.org/10.1093/gerona/qlr081>
- Vallabhajosula S, Roberts BL, Hass CJ (2014). Tai chi intervention improves dynamic postural control during gait initiation in older adults: a pilot study. *J Appl Biomech* 30:697–706. <https://doi.org/10.1123/jab.2013-0256>
- van Dieën JH, Spanjaard M, Könemann R, Bron L, Pijnappels M (2008). Mechanics of toe and heel landing in stepping down in ongoing gait. *J Biomech* 41:2417–2421. <https://doi.org/10.1016/j.jbiomech.2008.05.022>
- Vandervoort AA, Chesworth BM, Cunningham DA, Paterson DH, Rechnitzer PA, Koval JJ (1992). Age and sex effects on mobility of the human ankle. *J Gerontol* 47:M17–21. <https://doi.org/10.1093/geronj/47.1.m17>
- Vernazza S, Cincera M, Pedotti A, Massion J (1996). Balance control during lateral arm raising in humans. *Neuroreport* 7:1543–1548. <https://doi.org/10.1097/00001756-199607080-00002>
- Vialleron T, Delafontaine A, Ditcharles S, Fourcade P, Yiou E (2020). Effects of stretching exercises on human gait: a systematic review and meta-analysis. *F1000Res* 13; 9: 984 Doi: 10.12688/f1000research.25570.2.
- Vialleron T, Delafontaine A, Millerioux I, Memari S, Fourcade P, Yiou E (2021). Acute effects of short-term stretching of the *triceps surae* on ankle mobility and gait initiation in patients with Parkinson's disease. *Clinical Biomechanics* 89: 105449. <https://doi.org/10.1016/j.clinbiomech.2021.105449>.

- Vieira MF, Sacco I de CN, Nora FG da SA, Rosenbaum D, Hentschel Lobo da Costa P (2015). Footwear and Foam Surface Alter Gait Initiation of Typical Subjects. *PLoS ONE* 10:e0135821. <https://doi.org/10.1371/journal.pone.0135821>
- Vinti M, Couillandre A, Thoumie P (2010). Does somatosensory loss induce adaptation of the gait initiation process? *Neuroscience Letters* 480:178–181. <https://doi.org/10.1016/j.neulet.2010.06.017>
- Vuillerme N, Nafati G (2007). How attentional focus on body sway affects postural control during quiet standing. *Psychol Res* 71:192–200. <https://doi.org/10.1007/s00426-005-0018-2>
- Wang P, Yang X, Yang Y, Yang L, Zhou Y, Liu C, Reinhardt JD, He C (2015). Effects of whole body vibration on pain, stiffness and physical functions in patients with knee osteoarthritis: a systematic review and meta-analysis. *Clin Rehabil* 29:939–951. <https://doi.org/10.1177/0269215514564895>
- Wang Y, Asaka T, Watanabe K (2013). Multi-muscle synergies in elderly individuals: preparation to a step made under the self-paced and reaction time instructions. *Exp Brain Res* 226:463–472. <https://doi.org/10.1007/s00221-013-3449-9>
- Welter ML, Do MC, Chastan N, Torny F, Bloch F, Tézenas du Montcel S, Agid Y (2007). Control of vertical components of gait during initiation of walking in normal adults and patients with progressive supranuclear palsy. *Gait Posture* 26:393–399. <https://doi.org/10.1016/j.gaitpost.2006.10.005>
- Winter D (1995). Human balance and posture control during standing and walking. *Gait & Posture* 3:193–214. [https://doi.org/10.1016/0966-6362\(96\)82849-9](https://doi.org/10.1016/0966-6362(96)82849-9)
- Winter DA, Patla AE, Frank JS (1990). Assessment of balance control in humans. *Med Prog Technol* 16:31–51
- Woollacott MH, Shumway-Cook A, Nashner LM (1986). Aging and posture control: changes in sensory organization and muscular coordination. *Int J Aging Hum Dev* 23(2):97–114. doi: 10.2190/VXN3-N3RT-54JB-X16X.
- Woollacott MH, Tang PF, (1997). Balance control during walking in the older adult: research and its implications. *Phys Ther* 77(6):646–60. doi: 10.1093/ptj/77.6.646.
- World Health Organization (2008). WHO global report on falls prevention in older age. World Health Organization, Geneva, Switzerland
- Wu Z, Jiang X, Zhong M, Shen B, Zhu J, Pan Y, Dong J, Xu P, Zhang W, Zhang L (2020). Wearable Sensors Measure Ankle Joint Changes of Patients with Parkinson's Disease before and after Acute Levodopa Challenge. *Parkinsons Dis* 2020:2976535. doi: 10.1155/2020/2976535.

- Yiou E, Artico R, Teyssedre CA, Labaune O, Fourcade P (2016a). Anticipatory Postural Control of Stability during Gait Initiation Over Obstacles of Different Height and Distance Made Under Reaction-Time and Self-Initiated Instructions. *Front Hum Neurosci* 10:449. <https://doi.org/10.3389/fnhum.2016.00449>
- Yiou E, Caderby T, Delafontaine A, Fourcade P, Honeine JL (2017). Balance control during gait initiation: State-of-the-art and research perspectives. *World J Orthop* 8:815–828. <https://doi.org/10.5312/wjo.v8.i11.815>
- Yiou E, Caderby T, Hussein T (2012a). Adaptability of anticipatory postural adjustments associated with voluntary movement. *World J Orthop* 3:75–86. <https://doi.org/10.5312/wjo.v3.i6.75>
- Yiou E, Deroche T, Do MC, Woodman T (2011a). Influence of fear of falling on anticipatory postural control of medio-lateral stability during rapid leg flexion. *Eur J Appl Physiol* 111:611–620. <https://doi.org/10.1007/s00421-010-1680-7>
- Yiou E, Do MC (2001b). In a complex sequential movement, what component of the motor program is improved with intensive practice, sequence timing or ensemble motor learning? *Exp Brain Res* Mar 137(2):197-204. doi: 10.1007/s002210000624.
- Yiou E, Fourcade P, Artico R, Caderby T (2016b). Influence of temporal pressure constraint on the biomechanical organization of gait initiation made with or without an obstacle to clear. *Exp Brain Res* 234:1363–1375. <https://doi.org/10.1007/s00221-015-4319-4>
- Yiou E, Hamaoui A, Le Bozec S (2007). Influence of base of support size on arm pointing performance and associated anticipatory postural adjustments. *Neurosci Lett* 423:29–34. <https://doi.org/10.1016/j.neulet.2007.06.034>
- Yiou E, Hussein T, Larue J (2012b). Influence of temporal pressure on anticipatory postural control of medio-lateral stability during rapid leg flexion. *Gait Posture* 35:494–499. <https://doi.org/10.1016/j.gaitpost.2011.11.015>
- Zaidell LN, Mileva KN, Sumners DP, Bowtell JL (2013). Experimental Evidence of the Tonic Vibration Reflex during Whole-Body Vibration of the Loaded and Unloaded Leg. *PLoS One* 8(12):e85247. <https://doi.org/10.1371/journal.pone.0085247>
- Zattara M, Bouisset S (1988). Posturo-kinetic organisation during the early phase of voluntary upper limb movement. 1. Normal subjects. *Journal of Neurology, Neurosurgery & Psychiatry* 51:956–965. <https://doi.org/10.1136/jnnp.51.7.956>
- Zhu Z, Yin M, Cui L, Zhang Y, Hou W, Li Y, Zhao H (2018). Aquatic obstacle training improves freezing of gait in Parkinson's disease patients: a randomized

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# Annexes

## **Annexe 1 : Ajustements posturaux anticipateurs lors de l'initiation de la marche chez les patients hémiplésiques**

Cette étude a donné lieu à la publication suivante :

Delafontaine A, **Vialleron T**, Hussein T, Yiou E, Honeine JL, Colnaghi S (2019). Anticipatory Postural Adjustments During Gait Initiation in Stroke Patients. *Front Neurol* 10: 352. Doi: 10.3389/fneur.2019.00352.



# Anticipatory Postural Adjustments During Gait Initiation in Stroke Patients

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Prior to gait initiation (GI), anticipatory postural adjustments (GI-APA) are activated in order to reorganize posture, favorably for gait. In healthy subjects, the center of pressure (CoP) is displaced backward during GI-APA, bilaterally by reducing soleus activities and activating the tibialis anterior (TA) muscles, and laterally in the direction of the leading leg, by activating hip abductors. In post-stroke hemiparetic patients, TA, soleus and hip abductor activities are impaired on the paretic side. Reduction in non-affected triceps surae activity can also be observed. These may result in a decreased ability to execute GI-APA and to generate propulsion forces during step execution. A systematic review was conducted to provide an overview of the reorganization which occurs in GI-APA following stroke as well as of the most effective strategies for tailoring gait-rehabilitation to these patients. Sixteen articles were included, providing gait data from a total of 220 patients. Stroke patients show a decrease in the TA activity associated with difficulties in silencing soleus muscle activity of the paretic leg, a decreased CoP shift, lower propulsive anterior forces and a longer preparatory phase. Regarding possible gait-rehabilitation strategies, the selected studies show that initiating gait with the paretic leg provides poor balance. The use of the non-paretic as the leading leg can be a useful exercise to stimulate the paretic postural muscles.

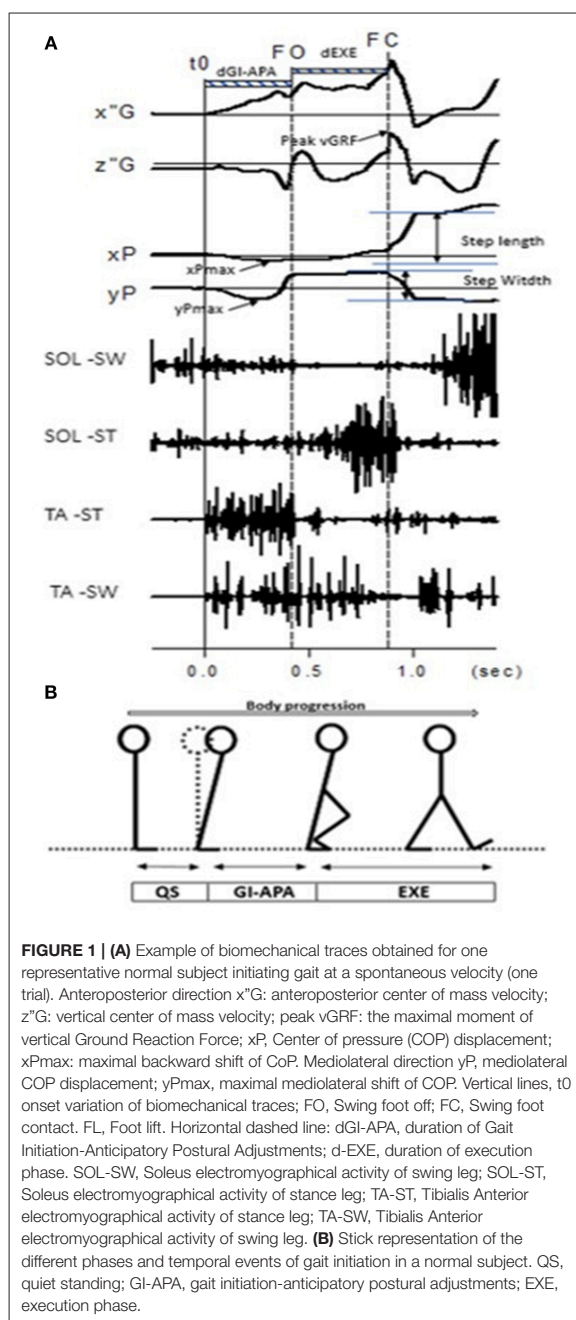
**Keywords:** anticipatory postural adjustments, gait initiation, stroke, rehabilitation, balance

## INTRODUCTION

Gait initiation (GI) is a functional task classically used in the literature to investigate balance control mechanisms in healthy and pathological individuals (1, 2). GI may be a challenging task for balance control in stroke patients because of their sensory and motor impairments (3–5). Better knowledge of the altered balance control mechanisms in these patients is important to allow clinicians to target rehabilitation programs directed at minimizing the risk of falling. However, to date there is no review involving the changes in balance control mechanisms during GI in stroke patients.

GI is composed of a postural phase (i.e., “anticipatory postural adjustments,” APA) where the dynamic phenomena necessary for a stable whole-body progression are generated, followed by the foot lift phase that ends at the time of swing toe clearance (i.e., body mass is transferred to the stance leg during this phase), and an execution phase that ends at the time of foot contact (2) (see **Figures 1A,B**).





Current literature describes gait initiation APA (GI-APA) components in the sagittal and frontal planes. In the sagittal plane, GI-APA are characterized by a backward center of pressure (CoP) shift with respect to the center of mass (CoM) position, providing the initial propulsive forces to progress forward (6, 7).

This CoP shift is caused mainly by a bilateral reduction in soleus tonic activity and an increase in tibialis anterior (TA) tonic activity (8). In the frontal plane, the CoP is displaced toward the leading leg during GI-APA, which serves to move the CoM in the contralateral direction. This CoM shift helps disengage the leading leg for step execution and is crucial for maintaining stability (9). This mediolateral CoP shift is thought to be caused by leading leg abduction, accompanied by trailing leg ankle dorsiflexion and knee flexion (10). In post-stroke hemiparetic patients, damage to central structures results in a reorganization of GI-APA. The goal of this systematic review was to provide an overview of the characteristics of GI-APA in this population.

## METHODS

### Methodology

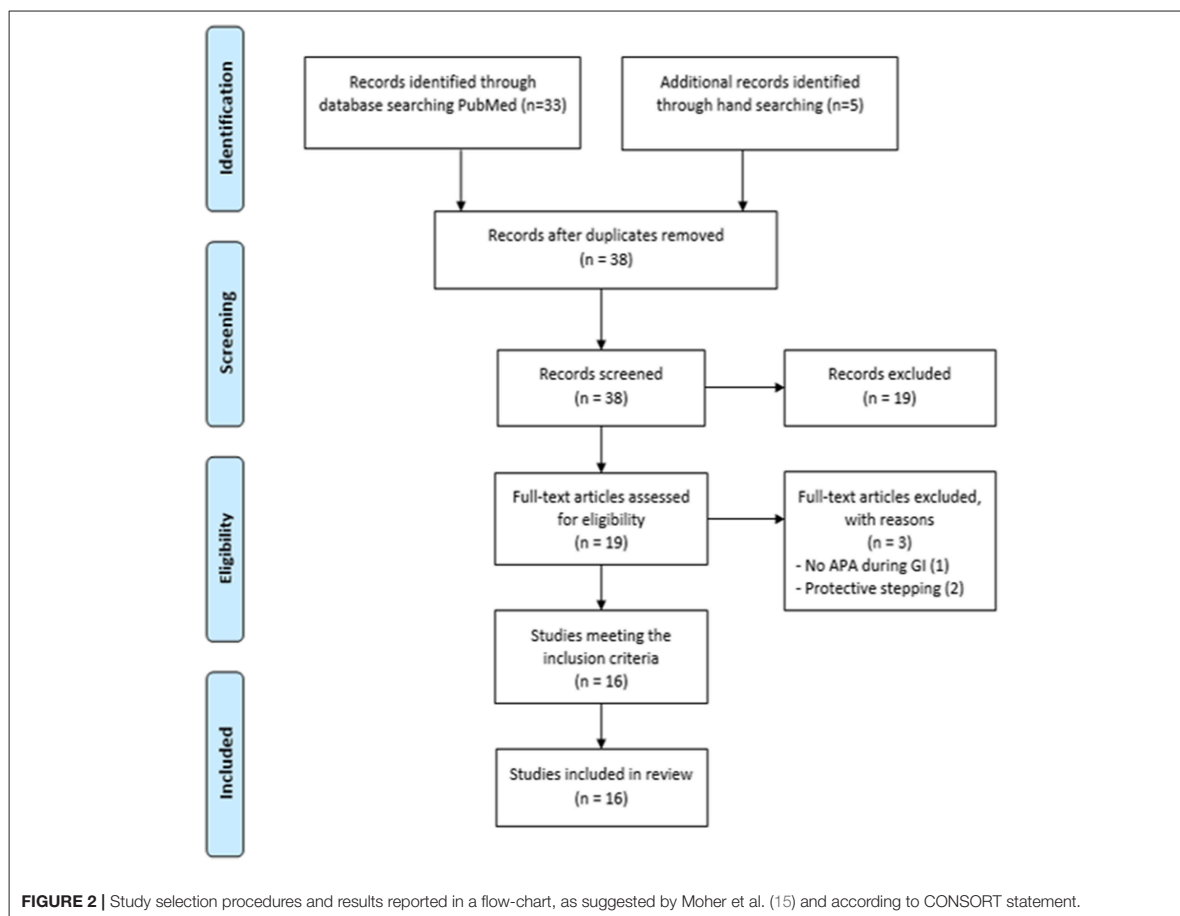
The Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) methodology was employed for this systematic review. A bibliographical search was conducted in PubMed and Google Scholar using the following keywords: (“gait initiation” OR “step initiation”) AND (“stroke” OR “hemiparesis” OR “hemiplegia”). Only articles available in English or French and published before 31 August 2018 were selected. Additionally, manual searches were conducted from the reference list of each article found in the database until no further citations could be identified. To be included, the studies had to meet all of the following inclusion criteria: (1) anticipatory postural adjustments were explored during the paradigm of gait initiation with a force platform or electromyographical (EMG) activity (6, 8); for review see (2), (2) at least one stroke patient group was studied, (3) all types of studies (i.e., randomized and non-randomized control trials), (4) only articles in English or French, and (5) articles published in peer-reviewed journals.

GI-APA were retained as a main inclusion criteria because it plays a major role in postural stability and GI performance [for review see (2)], notably for stroke patients which are able to develop different motor strategies regarding the leg which initiates the gait [i.e., paretic or non-paretic leg; (3–5, 11, 12)].

On the contrary, studies were excluded if they observed gait without the study of anticipatory postural adjustments or if no stroke patients were included in the protocol.

### Study Selection

Study selection was conducted in three steps (13, 14). First, two reviewers independently applied the inclusion and exclusion criteria to all the citations. Then, abstracts were screened to identify the titles representing a “best fit” with the aim of the present study. Finally, the reviewers read the full articles to reach a final decision on whether they should be included in the systematic review. When the relevance of a study was unclear from the abstract, the full article was read, and disagreements were resolved through group discussions with a third expert until a mutual consensus was reached. The reviewers met at the beginning, midpoint and final stages of the review process to discuss challenges and uncertainties related to study



selection, and to go back and refine the search strategy if needed (Figure 2).

### Quality Assessment

The Downs and Black scale was used to assess the risk of bias and the methodological quality in the selected studies. The scale was used for its ability to provide an overview of the study quality of not only randomized controlled trials but also non-randomized studies (16). Additionally, this tool has been demonstrated to have a significant correlation with the valid PEDro scale (17, 18). The Downs and Black scale are divided into 27 questions and the answer to each can be “yes” (one point), “no” or “unable to determine” (zero point). Only the last item was modified from the original scale, as described previously by Trac et al. (16). The maximum score is 28: one point for each question, except for question five (two points).

After all of the articles were assessed and total quality scores were calculated, the level of evidence appraisal was conducted using Grades of Recommendation, Assessment, Development, and Evaluation-Confidence (GRADE) as recommended by The Cochrane Qualitative and Implementation Methods Group (19).

The GRADE system classifies the quality of evidence in one of four levels: high, moderate, low, and very low. Evidence based on RCT begins as “high-quality,” whereas observational studies start with a “low quality” rating. Grading upwards may be warranted if the magnitude of the treatment effect is very large, if there is evidence of a dose-response relation or if all plausible biases would decrease the magnitude of an apparent treatment effect (20).

### Data Collection

Key items were charted from the selected articles by one of the authors. Item extraction was then checked by another author and disagreements were resolved by a third. The following data were noted for each selected article (Table 1):

- Author(s) and year of publication
- Study design
- Time elapsed between stroke onset and participation in the study
- Number of patients
- Mean age of patients

**TABLE 1** | Clinical characteristics of the population in the selected studies.

References	Post-stroke interval	Neurological lesion site	Symptoms	Symptoms severity	Treatment hints
Gama et al. (21)	6 months	NS	Hemiparetic gait	NS	NS
Rajachandrakumar et al. (12)	43 days	NS	Hemiparesis	CMSA leg: $4.9 \pm 1.3$ CMSA foot: $4.6 \pm 1.3$	NS
Sharma et al. (22)	NS	NS	Hemiparesis	FMA: score NS	NS
Sousa et al. (23)	$26 \pm 11.3$ months	Middle cerebral artery	Hemiparesis	FMA: score NS	NS
Sousa et al. (24)	$24.9 \pm 11.5$ months	internal capsule	Hemiparesis	FMA: score NS	NS
Martinez et al. (25)	$2.9 \pm 1.1$ yo	NS	Hemiparesis	NS	Use of stepping for balance rehabilitation
Ko et al. (26)	$5.2 \pm 3.1$ yo	Cerebrum for 11 patients and brainstem for 1 patient	Hemiparesis	FMA: score NS	NS
Chang et al. (27)	NS	8 PMC lesions, 7 PMC spared	Hemiplegia	FMA motor part: $30.1 \pm 2.5$ for PMC group; $32.4 \pm 1.1$ for PMC spared group.	NS
Melzer et al. (28)	$6.7 \pm 4.1$ yo	NS	Hemiparesis	BBS scale: $46 \pm 4.5$ for stroke patients	NS
Melzer et al. (29)	7.3 yo	Middle cerebral artery territory	Hemiparesis	BBS scale: $47.9 \pm 6.1$ on paretic side and $50.8 \pm 4.3$ on non-paretic side	NS
Bensoussan et al. (5)	7,66 yo (min 4, max 11)	Thalamic, internal capsule, sylvian area	Hemiplegia and equinus foot	FIM: 119–124 Barthel index: 95–100 Ashworth: 3–4	NS
Tokuno and Eng (30)	NS (i.e. minimum of 6 months post-stroke)	NS	Hemiparesis	CMSA: 8/14	Improve the propulsive force of the paretic leg when used as the trailing leg
Bensoussan et al. (4)	11 yo	Thalamic	Hemiplegia and equinus foot	Ashworth: 3/5 Barthel index: 95/100 FIM: 124–126 Functional ambulatory classification: 5/6	NS
Kirker et al. (31)	17 months	NS	Hemiparesis	NS	NS
Hesse et al. (11)	3.7 months	First ischemic stroke in middle cerebral artery	Hemiparesis	NS	NS
Brunt et al. (3)	11 weeks (min 4, max 23)	NS	Hemiplegia	NS	Improve symmetry in leg loading during quiet stance, or leg loading characteristics that are specific for gait initiation

CMSA, Chedoke-McMaster Stroke Assessment; FMA, Fugl-Meyer motor Assessment; FIM, Functional Independence measure; NS, Non-Specified; PMC, PreMotor Cortex; vs., versus.

- Lesion sites
- Symptoms and signs (extent and severity)
- Gait initiation paradigm
- APA measurement methods (force platform and/or EMG)
- APA features.

## RESULTS

### Selected Articles

Thirty-three articles emerged from the PubMed search. Five additional articles were identified through manual searching, for a total of 38 articles, 16 of which met the inclusion criteria (3–5, 11, 12, 21–23, 25–31). A total of 220 patients were included (Table 2).

### Patient Characteristics

The patients' age ranged between 21 and 86 years. In 11 of the 15 selected articles, GI was compared in healthy adults vs. stroke patients (5, 11, 21–24, 27–31). In the five remaining articles, only stroke patients were included (3, 4, 22, 25, 26). In 11 articles, participants had experienced stroke at least six months prior to GI recording (4, 5, 21, 23–26, 28–31). In one study, patients had experienced stroke 3.7 months prior to GI recording (11). In the remaining four studies, the post-stroke interval was not indicated (3, 12, 22, 27). The symptoms always consisted in hemiparesis, although its degree was rarely specified (4, 5, 25, 27–29). The presence and degree of spasticity was acknowledged only in Bensoussan et al. (4).

**TABLE 2** | Methods and main results of the selected studies.

Authors	Study design: (1) Main objectives (2) Study population (3) Outcomes (4) Methodology	Main results (only results related to GI-APA are reported)
Gama et al. (21)	<p>(1) First experiment: to compare gait initiation of individuals with stroke to age-matched non-disabled controls. Second experiment: to investigate how different amounts of partial body weight support (BWS) would influence the gait initiation of individuals with stroke.</p> <p>(2) 12 individuals with stroke (mean age <math>62 \pm 4.6</math> years, 3 women and 9 men) and 12 non-disabled controls (mean age <math>62 \pm 5</math> years, 9 women, 3 men).</p> <p>(3) Initial weight loading on swing and stance limbs and ML distance between heels, displacement and velocity in ML and AP directions of the CoP trajectories, step length and velocity of swing limb. With a force plate (model 9286BA, Kistler) and a computerized gait analysis system (VICON, Inc.) with 7 infrared cameras.</p> <p>(4) First experiment: after a verbal command, the participants were instructed to walk barefoot toward the end of a 7-m walkway at a comfortable speed with no interruption. Individuals with stroke started walking with the paretic limb, and individuals without disabilities started walking with the right limb. Second experiment: same instructions but 3 different conditions of body weight unloading were employed (0, 15, and 30%) in a randomized order, only with stroke patients.</p>	<p>Experiment 1: individuals with stroke presented shorter ML and AP CoP displacements backward and toward the swing limb during the anticipatory phase. Individuals with stroke presented slower velocity of the CoP in both ML and AP directions during the anticipatory phase.</p> <p>Experiment 2: no effect of the partial body support on the anticipatory phase.</p>
Rajachandrakumar et al. (12)	<p>(1) To determine the prevalence, clinical correlates, and consequences of atypical GI-APA (absent and multiple GI-APA) post-stroke when initiating gait with the paretic and non-paretic legs.</p> <p>(2) 40 individuals with sub-acute stroke able to initiate gait without using a gait aid (mean age 67.8 years; 19 men, and 21 women).</p> <p>(3) Temporal and spatial characteristics of the GI-APA (ML COP and CoM shifts) with a force plate.</p> <p>(4) 6 trials of gait initiation. Participants were instructed to step with a specific leg prior to the trial, and to initiate forward gait at a self-selected speed when ready after a verbal "go" signal.</p>	<p>35% of all trials had atypical GI-APA and most participants had at least one trial with an atypical GI-APA.</p> <p>Prevalence of atypical GI-APA in individuals with stroke was significantly related to motor impairment.</p> <p>Absent or multiple GI-APA when starting with the non-paretic leg.</p>
Sharma et al. (22)	<p>(1) To characterize ground reaction forces (GRF) acting on the legs during GI after stroke to begin to build an understanding of how hemiparesis might affect GI.</p> <p>(2) 18 chronic stroke patients (9 men, 3 women; mean age 67.6 years, range 45–86). 9 patients left paretic and 9 patients right paretic. Twenty eight age similar healthy subjects (mean age 67.6, range 49–82), 14 men and 14 women.</p> <p>(3) GRF was recorded from a Dual-Top Accusway force plate (AMTI, USA).</p> <p>(4) Subjects were instructed to initiate gait at the sound of an auditory cue. They self-selected their preferred speed. Left and right trials were collected in a random order and six trials were collected with each leg as the starting leg.</p>	<p>The legs of stroke patients generated lower anterior forces regardless of whether the trailing leg was paretic or non-paretic. Healthy subjects generated greater medial forces with their left leg than their right, but medial forces were equivalent for paretic and non-paretic legs. Patients with left hemisphere lesions generated greater lateral GRF with the non-paretic trailing leg than the paretic leg, and these non-paretic left leg lateral forces were also greater than the lateral forces generated by the left leg of healthy subjects. This left foot lateral GRF bias was not evident for patients with right hemisphere lesions.</p>
Sousa et al. (23)	<p>(1) To evaluate ankle GI-APA during gait initiation in chronic post-stroke subjects with lesion in the territory of middle cerebral artery (MCA).</p> <p>(2) 11 patients who had suffered a stroke at least 6 months earlier (mean age 53.10 years; 6 females, 5 males) and 12 healthy subjects (mean age 44.8 years; 5 females, 7 males).</p> <p>(3) Vertical, ML an AP ground reaction force with a force plate. The activity of Gastrocnemius Medialis (GM), Soleus (SOL) and Tibialis Anterior (TA) of both lower legs was assessed with EMG.</p> <p>(4) Subjects were asked to walk at self-adopted speed over a 5-m walkway, without explicit instructions. All participants from post-stroke group initiated gait with their contralesional leg.</p>	<p>Chronic post-stroke subjects present bilateral EMG SOL deactivation and lower tibialis anterior activity amplitude on the non- paretic leg and onset timing on both legs during gait initiation. This leads to decreased CoP displacement backward and toward the first swing leg.</p>

(Continued)

TABLE 2 | Continued

Authors	Study design: (1) Main objectives (2) Study population (3) Outcomes (4) Methodology	Main results (only results related to GI-APA are reported)
Sousa et al. (24)	<p>(1) To compare the reliability of CoP displacements during the postural phase of gait initiation calculated by two methods of detection the beginning of the postural phase in health and post-stroke participants.</p> <p>(2) 15 patients who had suffered a stroke at least 6 months earlier (8 females, 7 males) and 23 healthy participants (12 females, 11 males).</p> <p>(3) Vertical, ML and AP ground reaction force, as well as the values of moments of force in the frontal and sagittal planes, were acquired using a force plate (Bertec Corporation, USA).</p> <p>(4) To walk at self-selected speed over a 5 m walkway, without explicit instructions. Participants were asked to keep the starting leg consistent over all three trials.</p>	Post-stroke participants present decreased CoP displacement during the postural phase of gait initiation and increased postural phase duration that impairs posture stability and motor performance.
Martinez et al. (25)	<p>(1) To examine the stepping performance during voluntary and waist-pull perturbation-induced step initiation in people with chronic stroke.</p> <p>(2) 10 participants poststroke, (4 men and 6 women; mean age 59.6 years). Chronicity of stroke averaged 2.9 years, with 80% having right-sided weakness and ischemic stroke.</p> <p>(3) Ground reaction forces were recorded with 2 force platforms Body segment displacements were recorded using an optoelectronic system with 6-camera motion analysis system.</p> <p>(4) Voluntary and perturbation-induced stepping was studied consecutively. Three voluntary steps forward as soon and as fast as they could in response to a light "go" cue that followed a warning light stimulus by a variable delay period. Three trials for steps initiated with the non-paretic leg, 3 trials with the paretic leg. For the perturbation condition, steps were induced by forward postural perturbations.</p>	<p>GI-APA occurred in 100% of the voluntary stepping trials.</p> <p>The total postural phase duration was longer for voluntary stepping than induced straight forward stepping .</p> <p>The longer voluntary postural phase duration is primarily due to the prolonged unloading component. GI-APA amplitude was not different between conditions.</p>
Ko et al. (26)	<p>(1) To examine the immediate effect of natural (asymmetrical) and symmetrical weight bearing on the temporal events of ground reaction forces (GRF) and on timing and amplitude of lower distal muscle activity during GI in persons with hemiparesis.</p> <p>(2) 12 subjects (mean age 60.7 years; 7 males, 5 females) with unilateral hemiparesis.</p> <p>(3) EMG recording of Tibialis Anterior (TA) and Soleus (Sol) of both lower extremities.</p> <p>(4) The subject performed the following conditions of leg-loading distribution prior to executing a step in a randomized order: GI with the paretic leg during natural (asymmetrical) weight bearing; GI with the non-paretic leg during natural weight bearing; GI with the paretic leg during symmetrical weight bearing; and GI with the non-paretic leg during symmetrical weight bearing. Subjects began walking at their comfortable speed when they saw a visual cue (light). They were asked to initiate gait as quickly as possible after the visual cue. Subjects completed 5 trials.</p>	<p>Initiating gait with the non-paretic (leading) leg, the paretic TA muscle (trailing) was activated at the normal relative timing of percent GI cycle. When initiating gait with the non-paretic leg, the amplitude of the TA muscle on the paretic (trailing) leg was significantly increased about 27 to 36% compared with the paretic leg as leading leg in both natural and symmetrical leg- loading conditions.</p> <p>Improvement of standing symmetry during GI did not induce any adequate muscle activity on the impaired TA muscle for the effective weight shift necessary during GI. Delayed onset and less amplitude of TA activity when initiating with the paretic leg in both symmetrical and asymmetrical weight bearing conditions.</p>
Chang et al. (27)	<p>(1) To investigate whether premotor cortex (PMC) lesions influence stepping leg selection and stepping-related GI-APA in patients with a PMC lesion following a stroke.</p> <p>(2) 15 hemi-paretic patients (8 with PMC lesions, mean age 66.6 years and 7 PMC spared, mean age 65.1 years) and 8 age- and sex-matched healthy adults.</p> <p>(3) Vertical ground reaction force with 2 force plates EMG muscle activity from bilateral TA muscles was recorded.</p> <p>(4) All subjects were asked to prepare to react to an upcoming "go" signal (a green circle). Each subject underwent 6 right- and 6 left-leg-stepping trials in each condition. All subjects stepped independently without using any device.</p>	The PMC lesions group exhibited the longest TA reaction time and contraction latency on both legs.

(Continued)

TABLE 2 | Continued

Authors	Study design: (1) Main objectives (2) Study population (3) Outcomes (4) Methodology	Main results (only results related to GI-APA are reported)
Melzer et al. (28)	<p>(1) To explore differences in voluntary step behavior, a motor task of critical importance to prevent a fall from occurring, using the involved and uninvolved legs and to identify which of the step phases—step initiation-, preparation- or swing-phase—are markers for increased stepping time and thus, risk of falling.</p> <p>(2) 7 male and 3 female stroke survivors (61.7 ± 10 years old) 6.7 ± 4.1 years post- stroke and 10 healthy age-sex matched controls (61.7 ± 7.6 years old).</p> <p>(3) Center of pressure and ground reaction force data during step execution trials were collected and sampled at a frequency of 100 Hz, using a portable Kistler 9,287 force platform.</p> <p>(4) 12 step execution trials were performed in randomized order, 3 forward and 3 backward, for each leg, 6 stepping trials with the involved leg and 6 stepping trials with the uninvolved leg; healthy control subjects were instructed to step with their dominant leg.</p>	<p>The time to AP peak forces in the preparation and swing phases during single-task stepping were 80 and 162% longer, respectively, in the non-paretic leg compared with healthy controls.</p> <p>Also, the time to AP peak forces in the preparation and swing phases during single-task condition were 73 and 331% longer using the paretic leg in both stepping conditions compared with healthy subjects.</p> <p>AP peak force was significantly lower in both involved and uninvolved legs of stroke survivors compared with healthy controls, especially during the preparation phase in single- and dual-task conditions.</p>
Melzer et al. (29)	<p>(1) To ask whether the ability to quickly step—a motor task of critical importance to prevent a fall from occurring—was different in a group of chronic stroke survivors and healthy age- and sex-matched controls.</p> <p>(2) 16 hemiparetic subjects (age range, 46–77 year) who were 7.3 years poststroke, and 16 healthy controls matched for age and sex (age range, 43–81 years).</p> <p>(3) CoP and ground reaction force data during step execution trials were measured with a portable Kistler 9,287 force platform.</p> <p>(4) 3 forward and 3 backward stepping trials were performed by the uninvolved foot only in a randomized order for each of the 2 task conditions (single and dual task).</p>	<p>The preparatory phase durations in the chronic stroke survivors were significantly longer during the single task (93%), and only 35% longer in dual-task condition. A longer step initiation time was observed in stroke patients.</p>
Bensoussan et al. (5)	<p>(1) To assess the kinetic and kinematic characteristics of gait initiation during the various gait initiation phases in hemiplegic patients after stroke.</p> <p>(2) 3 hemiplegic patients after stroke (2 women, 1 man, 55, 35, and 21 years) were compared with 3 healthy subjects.</p> <p>(3) The kinematic study was carried out using an ELITE optoelectronic system (BTS spa, Milano, Italy) with 6 cameras The kinetic parameters were recorded via 2 AMTI force-plates (Advanced Mechanical Technology Inc., Watertown MA, USA).</p> <p>(4) The subject began walking at whatever speed he chose when instructed to do so by the operator. The foot to be used to initiate walking was chosen at random and specified by the operator by raising his left or right hand. Each subject performed 10 trials with each lower leg (20 trials/subjects).</p>	<p>The postural phase was longer when gait was initiated with the hemiplegic leg. The forces exerted by the affected leg were exerted backwards during the postural phase.</p>
Tokuno and Eng (30)	<p>(1) To determine the differences in gait initiation between a group of individuals with chronic stroke compared to a group of healthy adults.</p> <p>(2) 13 individuals with chronic stroke, 6 months post-stroke (8 males, 5 females, 59.3 years) and 13 healthy adults (8 males, 5 females, 64.7 years).</p> <p>(3) Gait initiation speed, duration of gait initiation, backward displacement of the center of pressure (CoP), initial loading of the legs during quiet stance, the peak vertical and antero-posterior (A-P) forces from the leading and trailing legs, and the A-P impulses from the leading and trailing leg with a force-plate.</p> <p>(4) Walking at their comfortable speed, without the use of any walking aid for a distance of five meters. The choice of leading leg was determined randomly by the experimenter. Three trials were collected from each lead leg condition.</p>	<p>Individuals with chronic stroke stood in a more asymmetrical manner. They relied only 42% on their paretic leg.</p> <p>The backward displacement of the CoP was smaller by a factor of approximately 2.5.</p> <p>The greater the ability an individual with stroke has to produce a larger peak A-P force and magnitude of A-P impulse with the paretic leg when used as the trailing leg, the faster the gait initiation speed they are able to achieve.</p> <p>Reduced propulsion by the paretic leg was observed.</p>

(Continued)

TABLE 2 | Continued

Authors	Study design: (1) Main objectives (2) Study population (3) Outcomes (4) Methodology	Main results (only results related to GI-APA are reported)
Bensoussan et al. (4)	(1) To investigate the temporal, kinetic and kinematic asymmetry of gait initiation in one subject with hemiplegia with an equinus varus foot. (2) 1 man with hemiplegia (21 years). (3) A kinetic analysis with two AMTI force plates and a kinematic analysis with an ELITE optoelectronic system. (4) The subject was asked to initiate gait at his preferred speed at a vocal signal. Operators randomly indicated to the subject which foot to use initiate gait. Ten trials for each foot	The distribution of body weight on the lower legs was asymmetrical. The forces exerted by the affected leg were exerted backwards during the postural phase. Reduced propulsion shortens monopodal phase and less body weight sharing by the paretic leg were observed.
Kirker et al. (31)	(1) To compare the pattern of pelvic girdle muscle activation in normal subjects and hemiparetic patients while stepping and maintaining standing balance. (2) 12 male and 5 female patients with stroke, with 13 right and 4 left hemiplegia's (mean age 54 years). Sixteen normal controls (mean age 46 years). (3) Surface EMGs of bilateral hip abductors (gluteus medius) and adductors (adductor longus), lateral torque component of the ground reaction forces, and pelvic displacement with a force-plate. (4) A sideways push task, a standing balance task and a gait initiation task. Forty trials were recorded, in alternate blocks of five trials leading with the left foot and five trials leading with the right foot.	The initial sideways weight shift is accompanied by increases in activity of the gluteus medius of the stepping leg and contralateral adductor. Lifting the weight off the leading leg is accompanied by large increases in activity of the trailing leg gluteus medius and leading leg adductor. Normal activation pattern of the hemiparetic leg muscles were observed.
Hesse et al. (11)	(1) To quantify a possible asymmetry in gait initiation of hemiparetic patients as compared to healthy subjects. (2) 14 hemiparetic patients (5 men and 9 women, mean age 62.8 years), mean stroke interval 3.7 months (range, 2.1–8.3). Five patients suffered from a right and 90 from a left hemiparesis. Ten healthy subjects (5 men, 5 women, mean age 57.3 years). (3) The vertical and horizontal forces were sampled with 2 triaxial force platforms. (4) After an acoustic signal, subjects started to walk at self-adopted speed, five times each with both their right and left leg.	Longer starting cycle duration and smaller backward displacement of the CoP in hemiparetic subjects compared to healthy subjects. The medio-lateral displacement of the CoP toward the trailing leg in the initial phase was less pronounced when the patient started with the paretic leg. When the patients made the first step with the paretic leg, the resulting trajectory of the CoM in the medio-lateral plane was very similar to that of healthy subjects. AP acceleration of the CoM was found to be delayed when starting with the affected leg.
Brunt et al. (3)	(1) To explore the relation between asymmetry in leg loading of persons with stroke and their ability to generate the appropriate forces to initiate gait; and to describe the relation between EMG activity of the involved and non-involved leg and the resultant ground reaction forces. (2) 13 patients with hemiplegia secondary to a cerebral vascular accident. (3) Surface electrodes were applied bilaterally to the center of the muscle belly of the medial gastrocnemius (G) and TA muscles. Vertical force before gait initiation was determined on a force-plate. (4) Walking at the onset of the light signal. The leading leg was always the involved leg. Six satisfactory trials in which the first three trials began with their involved leg on the force platform and the last three trials with their uninvolved leg on the force platform.	. The greater the leading leg was loaded before movement onset the more these forces approached those values observed in healthy adults. Reduced propulsion by the paretic leg. Correlation between symmetrical weight bearing and propulsion were observed. Decreased TA muscle activity clearly resulted in a decrease in the contribution of the leading leg to forward momentum before toe-off of that leg. There was often an increased G activity of the non-involved trailing leg that at times resulted in a TA and G co-contraction.

Only results related to GI-APA are reported. GI-APA, anticipatory postural adjustments; COP, center of pressure; COM, center of mass; ML, medio-lateral; AP, antero-posterior; TA, tibialis anterior; G, gastrocnemius.

## Task and Recordings

In four articles, participants were asked to perform a single step in a reaction-time paradigm, i.e., after the deliverance of a GO signal (25, 27–29). In the other articles, participants initiated gait spontaneously and continued walking at their own speed. In two articles (21, 23), the contralesional leg was systematically used as the leading leg. In the other articles, both legs were used as the leading leg.

Dynamic investigation using force platforms was conducted in all articles except one (26). EMG investigation was also conducted in five articles (3, 23, 26, 27, 31).

## Quality Assessment

The results of the quality assessments for each study are given in **Table 3**. Only non-randomized studies corresponded to the inclusion criteria and were evaluated. According to the modified Downs and Black scale, the mean quality index score for the studies was  $14 \pm 1.6$ , with scores ranging from 11 (3) to 17 (26, 29). Only non-randomized observational studies corresponded to the inclusion criteria and were evaluated, so the level of evidence was classified as “low quality.” No treatment was administered in the studies and no meta-analysis was conducted, so no grading upwards could be done.

## Biomechanical and EMG Features of APA: Stroke vs. Healthy Patients

### Biomechanics

#### APA pattern

Rajachandrakumar et al. (12) showed that when participants with sub-acute stroke initiated gait at their own speed, 78% exhibited at least one trial with a GI pattern classified as “atypical.” Atypical GI trial corresponded to a trial performed without APA or with “multiple” APA (i.e., with a multi-peaked CoP trace). In contrast, Martinez et al. (25) observed GI-APA in all trials that involved voluntary stepping in post-stroke patients (mean chronicity: 2.9 years), while APA were absent in 25% of trials involving perturbation-induced stepping.

#### APA duration

Rajachandrakumar et al. (12) showed that trials with multiple APA had a longer APA duration than normal GI trials (i.e., GI with a single peak CoP trace). A longer APA duration in stroke subjects compared to healthy subjects was reported by many authors (11, 23, 25, 27, 28). In addition, Gama et al. (21) recently established that stroke patients presented lower velocity of the CoP in both the ML and AP directions during the anticipatory phase of gait initiation.

#### Center-of-pressure shift

Sousa et al. (24) showed that post-stroke patients presented only half of the COP displacements (both in the frontal and sagittal plane) compared to healthy subjects. Gama et al. (21) demonstrated shorter ML and AP CoP displacements as well. A smaller COP backward shift was also demonstrated by Hesse et al. (11).

### Ground reaction forces

Melzer et al. (28) and Sharma et al. (22) showed that anteroposterior ground reaction forces generated by both the paretic and non-paretic leg in post-stroke patients during APA were lower compared to healthy controls.

Altogether, these studies indicate that stroke patients present impaired APA compared to healthy subjects, characterized by atypical patterns, longer duration and lower amplitude. The results reported in the paragraph below show that GI parameters can also differ depending on which leg (i.e., paretic or non-paretic) is used as the leading leg.

### Electromyography

Impairments of muscle activation during APA were demonstrated by many authors in stroke patients compared to healthy subjects (3, 23, 26, 31). Sousa et al. (24) showed that when the paretic leg was used as the leading leg, the TA of the paretic leg presented only half of the EMG activity observed in healthy subjects. On the other hand, Ko et al. (26) showed that when gait was initiated with the non-paretic leg, the TA activation in the paretic leg was increased by 27 to 36% compared with the paretic leg as leading leg. Moreover, stroke patients exhibited a bilateral decrease in TA activity during GI-APA and had difficulties in reducing tonic activity in the soleus muscle of the paretic leg (3). Finally, Kirker et al. (31) showed that when stroke patients initiated gait with the paretic leg, GI-APA were characterized by an impaired activation of the gluteus medius muscle of the paretic leg, which was compensated by more prominent activity in the contralateral adductor.

## APA in Stroke Patients: Paretic vs. Non-paretic leg as the Leading or Trailing leg

### APA Pattern

Rajachandrakumar et al. (12) showed that the frequency of trials with no APA did not differ when gait was initiated with the paretic or non-paretic leg. In contrast, multiple APA were more prevalent when the non-paretic leg was used as the leading leg. Additionally, when stepping with the non-paretic leg, the frequency of trials with no APA was negatively correlated with motor recovery.

### APA Duration

Bensoussan et al. (5) showed that the postural phase was longer when the paretic leg was the trailing leg. In contrast, Hesse et al. (11) and Melzer et al. (28) found no significant difference of postural phase duration when initiating gait with the paretic or non-paretic leg.

### APA Amplitude in the Frontal Plane

Sharma et al. (22) showed that patients with left hemisphere lesions generated greater lateral ground reaction forces with the non-paretic left leg than with the paretic right leg. In contrast, medial ground reaction forces were equivalent for both the paretic and non-paretic leg. Additionally, Hesse et al. (11) reported that the amplitude of the mediolateral CoP shift during APA was lower when gait was initiated with the paretic leg. In



**TABLE 3** | Quality assessment analysis according to the modified Downs and Black scale.

Authors	Items																											Total
	Reporting							External validity					Internal validity - bias							Internal validity—confounding					Power			
	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21	22	23	24	25	26	27	
Gama et al. (21)	1	1	1	1	0	1	1	0	1	1	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	14	
Rajachandrakumar et al. (12)	1	1	1	1	0	1	1	0	1	1	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	14	
Sharma et al. (22)	1	1	1	1	0	1	1	0	1	1	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	14	
Sousa et al. (23)	1	1	1	1	0	1	1	0	1	1	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	14	
Sousa et al. (24)	1	1	1	1	0	1	1	0	1	1	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	14	
Martinez et al. (25)	1	1	0	1	0	1	1	0	1	1	1	1	0	0	1	1	1	1	1	1	1	0	0	0	0	1	0	16
Ko et al. (26)	1	1	1	1	0	1	1	0	1	1	1	1	0	0	1	1	1	1	1	1	1	0	0	0	1	0	17	
Chang et al. (27)	1	1	1	1	0	1	1	0	1	1	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	14	
Melzer et al. (28)	1	1	1	1	0	1	1	0	1	1	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	1	15	
Melzer et al. (29)	1	1	1	1	0	1	1	0	1	1	1	0	0	0	1	1	1	1	1	1	1	0	0	0	1	1	17	
Bensoussan et al. (5)	1	1	1	1	0	1	1	0	1	0	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	13	
Tokuno and Eng (30)	1	1	1	1	0	1	1	0	1	0	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	13	
Bensoussan et al. (4)	1	1	1	1	0	1	1	0	1	1	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	14	
Kirker et al. (31)	1	1	0	1	0	1	1	0	1	0	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	12	
Hesse et al. (11)	1	1	1	1	0	1	1	0	1	0	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	13	
Brunt et al. (3)	1	1	0	1	0	1	0	0	1	0	0	0	0	0	1	1	1	1	1	1	0	0	0	0	1	0	11	

*Biomechanical and EMG features of APA: Stroke vs. healthy patients.*

contrast, the CoM did not move toward the trailing leg when patients started with their non-paretic leg, while the trajectory of the CoM was similar to that of healthy subjects when patients initiated gait with the paretic leg.

### APA Amplitude in the Sagittal Plane

Regarding APA in the sagittal plane, the same authors found that anticipatory anterior-posterior CoM acceleration during APA was lower when starting with the paretic leg (22). Tokuno and Eng (30) found that the anteroposterior impulse generated by the trailing leg was less than half that of healthy subjects. Additionally, Bensoussan et al. (4, 5) found that the ground reaction forces exerted by the paretic leg were directed backwards during APA (retropulsive action) but were directed forward for the non-paretic leg (propulsive action). Altogether, these findings indicate that the use of the paretic leg as the leading leg is associated with difficulties activating the TA and gluteus medius muscles during APA, which result in reduced CoP shift and CoM acceleration. In contrast, the use of the paretic leg as the trailing leg seems to challenge balance to a greater extent during GI.

## DISCUSSION

The aim of this systematic review was to provide an overview of the impact of stroke on the organization of APA associated with gait initiation (GI-APA). Altogether, the reviewed studies suggest that the anticipatory phase of GI is impaired in stroke patients. The findings principally highlight the changes in the pattern and spatiotemporal parameters of GI-APA in comparison with healthy subjects and the differences in initiating gait with the paretic or non-paretic leg. Moreover, this review may provide an evidenced-based theoretical framework for clinicians involved in rehabilitation programs for stroke patients.

The majority of the included studies were scored  $\geq 14/28$  according to the modified Downs and Black scale. The most common deficiencies in these included studies, in reference to the modified Downs and Black scale, were the lists of the principal confounders (question 5, i.e., not applicable for this review, only for interventional studies) and the adverse effects (question 8, i.e., not applicable for this review, only for interventional studies), the interventions not representative of usual care (question 13, i.e., not applicable for this review, only for interventional studies), the blinding of the patients or the assessors (questions 14 and 15, i.e., not applicable for this review, only for interventional studies), the specification of the time period of recruitment (question 22), the randomization of the subjects in groups (questions 23 and 24, i.e., not applicable for this review, only for randomized control trial studies) and the investigation of the main confounders (question 25, i.e., not applicable for this review, only for interventional studies). These absences were not surprising because the studies did not include intervention and control groups to measure the effects of treatment, but were observational studies comparing GI between subjects during a single session. Because of the absence of randomized controlled trials, a low quality level of evidence for the results may be provided (20). This means that further research with randomized protocols and more standardized protocols is very likely to have an important impact on our confidence in

the results. The level of the quality assessment, according to the modified Downs and Black scale, must be considered with caution because there is no treatment group (i.e., interventional randomized controlled trials) in the included studies.

Martinez et al. (25) found that when stroke patients initiated gait, GI-APA were present in 100% of trials, whereas Rajachandrakumar et al. (12) reported the possibility of trials with no GI-APA. The subjects in these two studies were at different post-stroke times and the significant changes in motor recovery occurring in the weeks after stroke may partially explain the observed differences (32, 33). Rajachandrakumar et al. (12) found that trials with no GI-APA were more prevalent among patients who walked slowly, and that motor impairment was correlated with the frequency of trials with no GI-APA. Multiple GI-APA, the other atypical pattern described by the authors, was observed at a higher frequency when gait was initiated with the healthy leg.

The “psychological apprehension” loading of the paretic leg may partially explain longer GI-APA duration and may reflect the priority of the central nervous system to secure mediolateral stability before walking (4, 5, 12). A longer GI-APA duration can also be attributed to delays in soleus deactivation (23) and TA activation (27). These muscle impairments probably lead to difficulties in correctly timing the CoP and CoM shift during GI-APA compared to healthy subjects.

Lower GI-APA amplitude observed in both the frontal and sagittal planes may result in lower stability and lower progression velocity, respectively, during GI (11, 21–23, 28, 30). However, the hypothesis that the lower gait progression velocity observed in stroke patients results from GI-APA impairments requires further investigation (23). Indeed, it has been shown that the amplitude of COP displacement is similar between healthy and post-stroke participants when healthy participants are instructed to walk at a speed similar to that chosen by the post-stroke subjects (30). Thus, the lower GI-APA amplitude reported in stroke patients under spontaneous velocity conditions appears to result from their lower self-adopted speed compared to healthy participants. Furthermore, balance and performance during the execution phase is affected by the choice of the leading leg during the postural phase. GI-APA are dynamic phenomena that could impact balance (34–36) and an inability to support body weight during GI due to weakness in the paretic leg. Thus, lower GI-APA amplitude in stroke patients could result from postural muscle weakness or reflect adaptations to the lower balance state.

Many studies included in the present review confirm the findings of Pélissier et al. (37) who described a reduction in the duration of the swing phase, step length, walking velocity, and magnitude of the propulsion force during the execution phase when initiating gait with the healthy leg. These effects could be the result of decreased plantar flexor, hip flexor and hip extensor strength on the paretic side (38, 39), which may underlie the inability to support body weight. The incapacity to correctly propel the CoM above the paretic leg during the stance phase may explain the higher degree of instability observed when starting with the non-paretic leg (3–5, 11, 22, 30, 31). Additionally, the paretic leg muscles contribute less to the load of the paretic side, generate less power, and may even produce retropulsive forces,

which may reduce walking velocity, because all of the propulsive forces are generated by the healthy leg alone (4, 5, 40).

Knowing the main characteristics of postural deficiencies during gait initiation or steady walking in stroke patients may help clinicians assess the efficiency of neurorehabilitation methods and provide an evidence-based theoretical framework for motor recovery. For example, the review of (37) established that the symmetry of stride was not correlated with walking performance (especially with the progression velocity). In addition, partial body support (i.e., on the non-paretic stance leg) was shown to have no effect on the anticipatory phase (21). These findings suggest that it is not necessary to obtain symmetrical static balance before starting gait rehabilitation. On the contrary, gait training should be started as early as possible after stroke. As documented in the literature reviewed above, initiating gait with the paretic leg improves stability and motor performance, so we recommend that rehabilitation programs promote this strategy in the early post-stroke period to improve patient mobility and autonomy.

Gamma et al. (21) recommend the use of an over-ground body weight support to improve mediolateral APA (i.e., stability) without changing the anteroposterior APA velocity (i.e., performance). However, in practice the combination of different rehabilitation strategies, notably the use of functional electrical stimulation and brain-computer interfaces, seems to be more effective than over-ground gait training alone (41).

Moreover, since using the paretic leg as the trailing leg has been found to increase the amplitude of TA muscle activity by between 27 and 36% (26), this strategy can be used as TA strength training to improve postural muscle performance in the paretic leg. Regarding possible gait-rehabilitation strategies, increasing TA muscle activity of the paretic leg during gait initiation seems to be an important target for motor rehabilitation, since bilateral TA activation during APA serves to provide initial propulsive

forces that shift the whole-body forward (3). It is also involved in body balance maintenance during step execution (10). It is clear that strength training should not be restricted to the TA of the paretic limb but should involve all paretic leg muscles. Specifically, reinforcing hip abductor muscles may facilitate the anticipatory CoP and CoM shifts in the frontal plane (31), while reinforcing hip extensors, knee extensors and plantar flexors should facilitate body weight support against gravity and provide efficient propulsive forces in the sagittal plane (5, 22, 28, 30).

In this way, the use of virtual reality training during gait initiation could help train and obtain recovery of force and power separately for each lower limb (i.e., paretic and non-paretic limb) in individuals with stroke (42).

## CONCLUSION

This systematic review provides an update on GI-APA reorganization following stroke. Stroke patients present atypical GI-APA patterns, longer GI-APA duration and lower GI-APA amplitude compared to healthy people, regardless of which leg is used as the leading or trailing leg. GI is facilitated when the non-paretic leg is used as the trailing leg because the weakness of the paretic leg leads to difficulties in supporting body weight during the upcoming stance phase. Further experiments should include distinct groups of patients in order to describe GI-APA features in acute, subacute and chronic stroke, and the influence of spasticity and of the lesion site. Understanding the changes in each population could be relevant for personalizing rehabilitation strategies.

## AUTHOR CONTRIBUTIONS

AD wrote the manuscript. TV and TH performed the bibliographical search and selected the articles to be included in the review. EY, J-LH, and SC reviewed the manuscript.

## REFERENCES

- Delafontaine A, Gagey O, Colnaghi S, Do M-C, Honeine J-L. Rigid ankle foot orthosis deteriorates mediolateral balance control and vertical braking during gait initiation. *Front Hum Neurosci.* (2017) 11:214. doi: 10.3389/fnhum.2017.00214
- Yiou E, Caderby T, Delafontaine A, Fourcade P, Honeine J-L. Balance control during gait initiation: State-of-the-art and research perspectives. *World J Orthoped.* (2017) 8:815–28. doi: 10.5312/wjo.v8.i11.815
- Brunt D, Vander Linden DW, Behrman AL. The Relation between limb loading and control parameters of gait initiation in persons with stroke. *Arch Phys Med Rehabil.* (1995) 76:627–34.
- Bensoussan L, Mesure S, Viton JM, Curvale G, Delarque A. Asymétries chronométriques, cinétiques et cinématiques de l'initiation de la marche chez un sujet hémiplégique. *Anna Réadapt Méd Physi.* (2004) 47:611–20. doi: 10.1016/S0168-6054(04)00177-1
- Bensoussan L, Mesure S, Viton J-M, Delarque A. Kinematic and kinetic asymmetries in hemiplegic patients' gait initiation patterns. *J Rehabil Med.* (2006) 38:287–94. doi: 10.1080/16501970600694859
- Lepers R, Brenière Y. The role of anticipatory postural adjustments and gravity in gait initiation. *Experimen Brain Res.* (1995) 107:118–24. doi: 10.1007/BF00228023
- Honeine J-L, Schieppati M, Gagey O, Do M-C. By counteracting gravity, triceps surae sets both kinematics and kinetics of gait. *Physiol Rep.* (2014) 2:e00229. doi: 10.1002/phy2.229
- Crenna P, Frigo C. A motor programme for the initiation of forward-oriented movements in humans. *J Physiol.* (1991) 437:635–53. doi: 10.1113/jphysiol.1991.sp018616
- Yiou E, Artico R, Teyssedre CA, Labaune O, Fourcade P. Anticipatory postural control of stability during gait initiation over obstacles of different height and distance made under reaction-time and self-initiated instructions. *Front Hum Neurosci.* (2016) 10:e00449. doi: 10.3389/fnhum.2016.00449
- Honeine J-L, Schieppati M, Crisafulli O, Do M-C. The neuro-mechanical processes that underlie goal-directed medio-lateral apa during gait initiation. *Front Hum Neurosci.* 10:445. doi: 10.3389/fnhum.2016.00445
- Hesse S, Reiter F, Jahnke M, Dawson M, Sarkodie-Gyan T, Mauri K-H. Asymmetry of gait initiation in hemiparetic stroke subjects. *Arch Phys Med Rehabil.* (1997) 78:719–24. doi: 10.1016/S0003-99939790079-4
- Rajachandrakumar R, Fraser JE, Schinkel-Ivy A, Inness EL, Biasin L, Brunton K, et al. Atypical anticipatory postural adjustments during gait initiation among individuals with sub-acute stroke. *Gait Posture.* (2017) 52:325–31. doi: 10.1016/j.gaitpost.2016.12.020
- Arksey H, O'Malley L. Scoping studies: towards a methodological framework. *Int J Soc Res Methodol.* (2005) 8:19–32. doi: 10.1080/1364557032000119616

14. Levac D, Colquhoun H, O'Brien KK. Scoping studies: advancing the methodology. *Implement Sci.* (2010) 5:69. doi: 10.1186/1748-5908-5-69
15. Moher D, Liberati A, Tetzlaff J, Altman DG. The PRISMA Group. preferred reporting items for systematic reviews and meta-analyses: the PRISMA statement *PLoS Med.* (2009) 6:e1000097. doi: 10.1371/journal.pmed.1000097
16. Trac MH, McArthur E, Jandoc R, Dixon SN, Nash DM, Hackam DG, et al. Macrolide antibiotics and the risk of ventricular arrhythmia in older adults. *Can Med Assoc J.* (2016) 188:E120–9. doi: 10.1503/cmaj.150901
17. de Morton NA. The PEDro scale is a valid measure of the methodological quality of clinical trials: a demographic study. *Aust J Physiother.* (2009) 55:129–33. doi: 10.1016/S0004-9514(09)70043-1
18. Aubut J-AL, Marshall S, Bayley M, Teasell RW. A comparison of the PEDro and Downs and Black quality assessment tools using the acquired brain injury intervention literature. *NeuroRehabilitation.* (2013) 32:95–102. doi: 10.3233/NRE-130826.
19. Noyes J, Booth A, Flemming K, Garside R, Harden A, Lewin S, et al. Cochrane qualitative and implementation methods group guidance series-paper 3: methods for assessing methodological limitations, data extraction and synthesis, and confidence in synthesized qualitative findings. *J Clin Epidemiol.* (2018) 97:49–58. doi: 10.1016/j.jclinepi.2017.06.020
20. Guyatt GH, Oxman AD, Vist GE, Kunz R, Falck-Ytter Y, Alonso-Coello P, et al. GRADE: an emerging consensus on rating quality of evidence and strength of recommendations. *Br Med J.* (2008) 336:924–6. doi: 10.1136/bmj.39489.470347.AD
21. Gama GL, Celestino ML, Barela JA, Barela AMF. Gait initiation and partial body weight unloading for functional improvement in post-stroke individuals. *Gait Posture.* (2019) 68:305–10. doi: 10.1016/j.gaitpost.2018.12.008
22. Sharma S, McMorland AJC, Stinear JW. Stance limb ground reaction forces in high functioning stroke and healthy subjects during gait initiation. *Clin Biomech.* (2015) 30:689–95. doi: 10.1016/j.clinbiomech.2015.05.004
23. Sousa A, Silva A, Santos R. Ankle anticipatory postural adjustments during gait initiation in healthy and post-stroke subjects. *Clin Biomech.* (2015) 30:960–5. doi: 10.1016/j.clinbiomech.2015.07.002
24. Sousa A, Silva A, Santos R. Reliability of two methods for identifying the postural phase of gait initiation in healthy and poststroke subjects. *J Appl Biomech.* (2015) 31:349–56. doi: 10.1123/jab.2014-0222
25. Martinez KM, Mille M-L, Zhang Y, Rogers MW. Stepping in persons poststroke: comparison of voluntary and perturbation-induced responses. *Arch Phys Med Rehabil.* (2013) 94:2425–32. doi: 10.1016/j.apmr.2013.06.030
26. Ko M, Bishop MD, Behrman AL. Effects of limb loading on gait initiation in persons with moderate hemiparesis. *Top Stroke Rehabil.* (2011) 18:258–68. doi: 10.1310/tsr1803-258
27. Chang W-H, Tang P-F, Wang Y-H, Lin K-H, Chiu M-J, Chen S-HA. Role of the premotor cortex in leg selection and anticipatory postural adjustments associated with a rapid stepping task in patients with stroke. *Gait Posture.* (2010) 32:487–93. doi: 10.1016/j.gaitpost.2010.07.007
28. Melzer I, Goldring M, Melzer Y, Green E, Tzedek I. Voluntary stepping behavior under single- and dual-task conditions in chronic stroke survivors: A comparison between the involved and uninvolved legs. *J Electromyogr Kinesiol.* (2010) 20:1082–7. doi: 10.1016/j.jelekin.2010.07.001
29. Melzer I, Tzedek I, Or M, Shvartsh G, Nizri O, Ben-Shitrit K, et al. Speed of voluntary stepping in chronic stroke survivors under single- and dual-task conditions: a case-control study. *Arch Phys Med Rehabil.* (2009) 90:927–33. doi: 10.1016/j.apmr.2008.12.012
30. Tokuno CD, Eng JJ. Gait initiation is dependent on the function of the paretic trailing limb in individuals with stroke. *Gait Posture.* (2006) 24:424–8. doi: 10.1016/j.gaitpost.2005.09.012
31. Kirker SGB, Simpson DS, Jenner JR, Miles Wing A. Stepping before standing: hip muscle function in stepping and standing balance after stroke. *J Neurol Neurosurg Psychiatry.* (2000) 68:458–64. doi: 10.1136/jnnp.68.4.458
32. Jorgensen HS, Nakayama H, Raaschou HO, Vive-Larsen J, Stoier M, Olsen TS. Outcome and time course of recovery in stroke. Part II: time course of recovery the Copenhagen Stroke Study *Arch Phys Med Rehabil.* (1995) 76:406–12.
33. Verheyden G, Nieuwboer A, De Wit L, Thijs V, Dobbelaere J, Devos H, et al. Time course of trunk, arm, leg, and functional recovery after ischemic stroke. *Neurorehabil. Neural Repair.* (2008) 22:173–9. doi: 10.1177/1545968307305456
34. Nouillot P, Bouisset S, Do MC. Do fast voluntary movements necessitate anticipatory postural adjustments even if equilibrium is unstable? *Neurosci Lett.* (1992) 147:1–4.
35. Aruin AS, Forrest WR, Latash ML. Anticipatory postural adjustments in conditions of postural instability. *Electroencephalogr Clin Neurophysiol.* (1998) 109:350–9. doi: 10.1016/S0924-980X(98)00029-0
36. Yiou E, Mezaour M, Le Bozec S. Anticipatory postural adjustments and focal performance during bilateral forward-reach task under different stance conditions. *Motor Control.* (2009) 13:142–60. doi: 10.1123/mcj.13.2.142
37. Pélissier J, Pérennou D, Laassel E. Analyse instrumentale de la marche de l'hémiplégique adulte: revue de la littérature. *Ann Réadapt Méd Phys.* (1997) 40:297–313.
38. Lindmark B, Hamrin E. Relation between gait speed, knee muscle torque and motor scores in post-stroke patients. *Scand J Caring Sci.* (1995) 9:195–202. doi: 10.1111/j.1471-6712.1995.tb00414.x
39. Nadeau S, Arseneault AB, Gravel D, Bourbonnais D. Analysis of the clinical factors determining natural and maximal gait speeds in adults with a stroke. *Am J Phys Med Rehabil.* (1999) 78:123–30. doi: 10.1097/00002060-199903000-00007
40. Peterson CL, Hall AL, Kautz SA, Neptune RR. Pre-swing deficits in forward propulsion, swing initiation and power generation by individual muscles during hemiparetic walking. *J Biomech.* (2010) 43:2348–55. doi: 10.1016/j.jbiomech.2010.04.027
41. Belda-Lois JM, Mena-del Horno S, Bermejo-Bosch I, Moreno JC, Pons JL, Farina D, et al. Rehabilitation of gait after stroke: a review towards a top-down approach. *J Neuroeng. Rehabil.* (2011) 8:66. doi: 10.1186/1743-0003-8-66
42. Mirelman A, Patrilli BL, Bonato, P, Deutsch, JE. (2010) Effects of virtual reality training on gait biomechanics of individuals post-stroke. *Gait and Posture.* 31, 433–437. doi: 10.1016/j.gaitpost.2010.01.016

**Conflict of Interest Statement:** The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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## Annexe 2 : Articles publiés et communications orales en congrès internationaux

### Articles publiés dans des revues internationales avec comité de relecture :

- 1) **Vialleron T**, Delafontaine A, Millerioux I, Memari S, Fourcade P, Yiou E (2021). Acute effects of short-term stretching of the *triceps surae* on ankle mobility and gait initiation in patients with Parkinson's disease. *Clinical Biomechanics* 89: 105449. <https://doi.org/10.1016/j.clinbiomech.2021.105449>.
- 2) **Vialleron T**, Delafontaine A, Ditcharles S, Fourcade P, Yiou E (2020). Effects of stretching exercises on human gait: a systematic review and meta-analysis. *F1000Res* 13; 9: 984 Doi: 10.12688/f1000research.25570.2.
- 3) Delafontaine A, **Vialleron T**, Fischer M, Laffaye G, Chèze L, Artico R, Genêt F, Fourcade P, Yiou E (2019). Acute Effects of Whole-Body Vibration on the Postural Organization of Gait Initiation in Young Adults and Elderly: A Randomized Sham Intervention Study. *Front Neurol* 10: 1023. Doi: 10.3389/fneur.2019.0102.
- 4) Fischer M, **Vialleron T**, Laffaye G, Fourcade P, Hussein T, Chèze L, Deleu PA, Honeine JL, Yiou E, Delafontaine A (2019). Long-Term Effects of Whole-Body Vibration on Human Gait: A Systematic Review and Meta-Analysis. *Front Neurol* 10: 627. Doi: 10.3389/fneur.2019.00627.
- 5) Delafontaine A, **Vialleron T**, Hussein T, Yiou E, Honeine JL, Colnaghi S (2019). Anticipatory Postural Adjustments During Gait Initiation in Stroke Patients. *Front Neurol* 10: 352. Doi: 10.3389/fneur.2019.00352.

### Communications orales en congrès internationaux

- 1) Delafontaine A, **Vialleron T**, Fischer M, Laffaye G, Chèze L, Artico R, Honeine JL, Fourcade P and Yiou E. Post-effet biomécanique immédiat des vibrations corps entier sur l'initiation de la marche des sujets âgés versus jeunes : étude

randomisée contrôlée avec placebo. iCEPS 2019, 28-30 mars 2019, Nouvelle faculté de médecine, Montpellier, France.

- 2) Delafontaine A, **Vialleron T**, Fischer M, Laffaye G, Chèze L, Artico R, Honeine JL, Fourcade P and Yiou E. Adaptation of anticipatory postural adjustments during gait initiation induced by short-term whole-body vibration: a randomized sham intervention study in young adults and elderly. 12th International Society of Physical and Rehabilitation Medicine World Congress (ISPRM), 8-12 Juillet 2018, Palais des Congrès, Paris.
- 3) Delafontaine A, **Vialleron T**, Fischer M, Laffaye G, Fourcade P and Yiou E. Immediate biomechanical effects of whole-body vibration on mediolateral balance control during gait initiation in young healthy adults and elderly: a randomized placebo study. 8th World Congress on Biomechanics (WCB), 8-12 Juillet 2018, Dublin, Ireland.

**Titre :** Analyse des effets d'une modification expérimentale de la capacité posturo-cinétique induite par deux types d'exercices physiques sur l'organisation biomécanique de la marche et de l'initiation de la marche : vibrations du corps entier et étirements musculaires.

**Mots clés :** initiation de la marche, marche, ajustements posturaux anticipateurs, capacité posturo-cinétique, vibrations corps entier, étirements musculaires, vieillissement, maladie de Parkinson, biomécanique.

**Résumé :** La « capacité posturo-cinétique » correspond à la capacité du système nerveux central à générer des ajustements posturaux en réponse à une perturbation de la posture et de l'équilibre. Cette capacité dépend étroitement de la mobilité articulaire. L'objectif de cette thèse était d'étudier les effets d'une modification de la capacité posturo-cinétique induite par deux types d'exercices physiques (les « vibrations corps entier » [VCE] et les étirements musculaires) sur l'organisation biomécanique de la marche et de l'initiation de la marche dans différents types de populations (saines et pathologiques). Pour chaque type d'exercice, nous avons réalisé, d'une part, une revue systématique de la littérature avec méta-analyse sur les effets à long terme sur la marche et, d'autre part, une étude expérimentale sur les effets à court terme sur l'initiation de la marche. Les populations incluses dans nos études expérimentales étaient des jeunes adultes et des personnes âgées saines (VCE) et des patients atteints de la maladie de Parkinson (étirements musculaires).

Les résultats de notre première expérimentation ont montré qu'une séance de VCE suffisait à améliorer la

stabilité des jeunes adultes, et ce, malgré une augmentation de la raideur du membre portant. Cette amélioration de la stabilité était rendue possible grâce à une augmentation de l'amplitude des « ajustements posturaux anticipateurs » (APA). Aucun effet à court terme des VCE n'a été observé sur la raideur et les paramètres de l'initiation de la marche chez les personnes âgées en bonne santé. Les résultats de notre méta-analyse ont cependant montré que l'exposition aux VCE améliorait la marche dans cette population sur le long terme. Les résultats de notre seconde expérimentation ont montré qu'une seule séance d'étirements du triceps sural permettait d'améliorer le développement des APA et la performance motrice chez des patients parkinsoniens. En revanche, notre méta-analyse n'a pu démontrer de bénéfice significatif des étirements sur le long terme dans cette population par manque de données dans la littérature. L'ensemble de ces résultats conforte le concept biomécanique de capacité posturo-cinétique et devrait avoir des répercussions pertinentes dans le domaine de la réhabilitation motrice.

**Title:** Effects of an experimental modification of the posturo-kinetic capacity induced by two types of physical exercises on the biomechanical organization of walking and gait initiation: whole-body vibrations and muscle stretching.

**Keywords:** gait initiation, walking, anticipatory postural adjustments, posturo-kinetic capacity, whole-body vibrations, muscle stretching, aging, Parkinson disease, biomechanics.

**Abstract:** "Posturo-kinetic capacity" refers to the central nervous system's ability to generate postural adjustments in response to a postural perturbation. This ability is known to depend on joint mobility. The objective of this thesis was to study the effects of a change in posturo-kinetic capacity induced by two types of physical exercise (whole-body vibrations [WBV] and muscle stretching) on the biomechanical organization of walking and gait initiation in different types of population (healthy and pathological). For each exercise, we conducted a systematic review of the literature with meta-analysis on long-term effects on walking, and an experimental study on short-term effects on gait initiation. The populations included in the experimental studies were healthy young adults and elderly (WBV) and patients with Parkinson's disease (muscle stretching).

Results of the first experiment showed that a single WBV session was enough to improve stability in young adults,

despite an increase in the supporting limb stiffness. This improvement was made possible by an increase in the amplitude of "anticipating postural adjustments" (APA). In healthy elderly, no short-term WBV effects were observed on leg stiffness and gait initiation parameters. However, results of the meta-analysis showed that exposure to WBV improved walking in this population over the long term. Results of our second experiment showed that a single session of triceps surae stretching improved APA development and motor performance in patients with Parkinson's disease. However, the meta-analysis could not demonstrate a significant benefit of long-term stretching on walking in this population due to a lack of data in the literature. Together, these results support the biomechanical concept of posturo-kinetic capacity and may have relevant applications in the field of motor rehabilitation.