Université de Montréal

Understanding the Underlying Biomechanical Mechanisms and Strategies in Dysvascular Lower-Limb Amputees during Gait Initiation: Implications for Gait Analysis

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Résumé

Un grand nombre de paramètres biomécaniques sont disponibles pour quantifier la marche mais aucun consensus n'existe quant aux paramètres les plus pertinents à mesurer lors de l'analyse de la marche chez les sujets sains. Le premier objectif de cette thèse était donc de réaliser une revue systématique afin de déterminer les paramètres les plus pertinents pour l'analyse de la marche chez les adultes sains. Les résultats ont permis de confirmer que les paramètres spatiaux-temporaux, et plus spécifiquement la vitesse de marche, sont les paramètres les plus souvent mesurés par le plus grand nombre d'articles pour l'analyse de la marche chez les adultes sains. De futures études sont nécessaires afin de pouvoir comparer ces résultats chez d'autres populations et déterminer leur pertinence clinique.

Lors de l'initiation à la marche, les ajustements posturaux anticipatoires (APA) permettent le transfert du poids du corps et la propulsion, tout en gardant l'équilibre et, au premier pas, de 75% à 90% de la vitesse de marche moyenne (SSWV) est atteinte. Bien que la population d'amputés transtibiaux pour cause dysvasculaire (DTTA) est importante et grandissante, aucune étude n'a, à ce jour, examiné les APA et le patron du premier pas lors de l'initiation à la marche chez cette population. Les deuxième et troisième objectifs de cette thèse étaient donc de comparer le patron des APA et la cinétique du premier pas chez 10 DTTA et 10 sujets contrôles lors de l'initiation à la marche. Les sujets ont initié la marche avec la jambe droite et gauche jusqu'à ce que la SSWV soit atteinte.

Les résultats de la deuxième étude démontrent une augmentation du temps en phase APA chez les DTTA, une stratégie pour compenser la force réduite en augmentant l'impulsion. Le résultat le plus important chez le DTTA est qu'en A/P, un déplacement total antérieur a été observé sous la jambe prothétique, une stratégie qui semble être spécifiquement associé à l'amputation dysvasculaire.

Les résultats de la troisième étude démontrent que lors du premier pas, l'impulsion de propulsion de la jambe prothétique était réduite par rapport à la jambe intacte et aux sujets contrôles. Cette réduction d'impulsion de propulsion est directement reliée à la perte des muscles fléchisseurs-plantaires au niveau de la jambe amputée. Curieusement, pour la force verticale maximale lors de la mise en charge et le taux de chargement, aucune différence n'a été observée entre la jambe intacte et la jambe des sujets contrôles ce qui supporte l'idée que les DTTA profitent d'un facteur protecteur contre le risque d'ostéo-arthrite au niveau de la jambe intacte.

Les spécialistes travaillant avec les DTTA devraient promouvoir l'initiation de la marche avec les deux jambes afin de bien préparer le DTTA aux perturbations de la vie quotidienne. Également, l'augmentation de la SSWV ne devrait pas nécessairement être un objectif de la réadaptation. De prochaines études devraient s'intéresser à comparer le patron de marche chez les DTTA aux amputées pour cause traumatique ainsi que s'intéresser au patron de terminaison de la marche.

Mots clés : ajustements posturaux anticipatoires, marche, initiation de la marche, biomécanique, amputé transtibial**.**

Abstract

A large number of biomechanical parameters are readily available with which to quantify gait but no consensus on the most relevant parameters for gait analysis in healthy adults exists with which to compare these results. The first objective of this thesis was therefore to complete a systematic review in order to establish those parameters most relevant for gait analysis in healthy adults. Results showed spatio-temporal parameters, specifically walking velocity, to be the most often measured biomechanical parameters and reported by the greatest number of articles for gait analysis in the healthy adult population. Further research should aim to compare these results to those of other populations and determine their clinical relevance.

In gait initiation, anticipatory postural adjustments (APA) allow for body weight to be transferred and propulsion while maintaining balance. As well, the first step accounts for 75% to 90% of the steady-state walking velocity (SSWV). Though the dysvascular transtibial amputee (DTTA) is the most sizeable and growing amputee population, no studies have yet investigated the APA's and first step gait initiation pattern in this specific population. Thus, the second and third objectives of this thesis were aimed at comparing the APA's pattern and underlying first step kinetics in 10 DTTA with 10 healthy controls prior during gait initiation. Participants were asked to initiate gait with their right then left limb leading until they reached SSWV.

In the second study, the increased APA time observed in the DTTA support the strategy to improve impulse by increasing time in the presence of diminished force production. The most important result is with regards to A/P total APA, as a total anterior displacement was

observed in the prosthetic limb and would appear to be related to further reductions in propulsion specifically associated with dysvascular amputation.

Lastly, the results of the third study showed that propulsive impulse was significantly reduced in the prosthetic limb when compared to intact and control limbs. The reduction in propulsive impulse testifies of the missing plantarflexor muscles of the prosthetic limb. Interestingly, with regards to maximum vertical force at weight acceptance and loading rate there was no difference between the intact limb and the control limb. Though the DTTA are able to produce less intact limb vertical force, this may also place them at a reduced osteoarthritis risk in the intact limb.

Rehabilitation specialists should focus on both prosthetic and intact leading limb for gait initiation to aid the DTTA with everyday perturbations. As well, increasing SSWV should perhaps not be a goal of rehabilitation. Future research should focus on comparing gait initiation in the DTTA when compared to the traumatic TTA counterpart as well as understanding gait termination in the DTTA.

Key words: anticipatory postural adjustments, gait, gait initiation, biomechanics, transtibial amputee.

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List of Abbreviations

APA	Anticipatory postural adjustment
A/P	Antero-posterior
BW	Body weight
CoM	Center of mass
CoP_{net}	Net center of pressure
DTTA	Dysvascular transtibial amputee
GRF	Ground reaction force
M/L	Medio-lateral
RMS	Root mean square
SSWV	Steady-state walking velocity
TTA	Transtibial amputee
3D	Three-dimensional

Chapter 1: **Introduction**

Because walking is the most common form of locomotion and is part of almost all activities of daily living [1,2], the ability to walk is an indicator of overall health and autonomy [1]. Although walking is usually learned at a young age, the mechanics of walking are quite complex and they have been the focus of numerous studies since the beginning of the 18th century.

From the first studies of human walking elaborated through a series of photographic images, by early biomechanics enthusiasts Edweard Muybridge and Étienne-Jules Marey, gait analysis as it is known today has evolved significantly [2]. Numerous studies have specifically investigated the walking pattern in healthy adults, especially as this is a sizeable population, and their walking pattern is today better understood [3,4,5,6,7,8]. The gait mechanics of healthy adults is often compared across ages (i.e. development, maturation and degeneration process) and with of other populations where gait disorders may be associated with deficits and pathologies [9,10,11]. Thus, all aspects of gait in healthy adults must be well understood in order to make appropriate recommendations for rehabilitation, footwear, walking aids, orthosis and prosthesis.

Indeed, the process of the walking activity can be broken down into various phases, from quiet standing to gait initiation, and finally to steady-state walking. Placing a focus on gait initiation and steady-state walking, the current thesis aims at comparing the anticipatory postural adjustment (APA) and the forward movement production during gait initiation up to steady-state walking velocity (SSWV) in both healthy and lower limb amputation populations.

In the following sections, pertinent biomechanical and clinical literatures, on how lower limb amputees and healthy adults are able to initiate walking from a quiet standing position and accelerate to reach SSWV, will be presented. But, first, the epidemiology (i.e. incidence, causes, level of amputation) and particularities of the lower limb amputee population will be presented.

1. Etiology of the Lower Limb Amputee Population

   1.1 Incidence of lower limb amputation

In the United States, the incidence of lower limb amputation is 45/100 000 habitants [12]. It is also estimated that over 1.6 million people are living with a lower limb amputation [13]. Lower extremity amputations affect mostly men (i.e. 69%) above the age of 50 to 60 years old [14,15,16].

Trends indicate that the number of amputations will continue to increase, in large part because of the increasing number of individuals affected by dysvascular diseases associated with diabetes [17]. Projections indicate that by 2050, the number of lower limb amputations associated with vascular problems will be over 2.27 million people in the United States [13].

For Canada, the estimation of the total number of lower limb amputees is not available, but Imam and colleagues (2017) have reported that an average of 7405 new amputations are performed every year with an incidence of 28.9/100 000 Canadians with diabetes [15].

Even when adjusted for age and/or sex in individuals with diabetes, important rates of lower limb amputation, as high as 176 and 158/100 000 are reported, respectively for the Republic of Ireland and France [18,19]. These high rates of lower limb amputation are those specifically calculated for the population who are affected by diabetes mellitus and its numerous complications. The next section will address in details the complications experienced by the lower limb amputee population.

1.2 Causes of lower limb amputation

Dysvascularity and diabetes represent the main causes in 76 to 96% of the total number of lower limb amputations in Canada and the United States [12,16]. As introduced above, diabetes and peripheral vascular disease are not only associated to but are also the major causes of lower limb amputations as type II diabetes is a principle cause of vascular problems and associated complications (i.e. hypertension, atherosclerosis, thrombosis, peripheral neuropathy, retinopathy, nephropathy and cataracts) which currently impacts over 30 million individuals in the United States, and this number is said to be increasing. It has also been estimated that over 7.4 million individuals in the United States are living with type II diabetes without knowing of their medical condition [20].

With increasing time since onset of type II diabetes, there is also an increased risk for developing associated complications such as peripheral neuropathy, the most common complication associated to diabetes [21,22]. Peripheral neuropathy results from damage of the nerves responsible for the transport of the information from pressure receptors of the sole of the foot to the brain. These patients lose sensitivity from the sole of their feet and are therefore unable to sense injuries such as blisters. Infections, if left untreated, can progress rapidly, leading to gangrene which follows suit to a lower limb amputation [23].

Other reasons for lower limb amputations are trauma (i.e. such as car and combat related accidents), cancer and congenital deformity or defects. In the United States, it has been surveyed that cancer and congenital reasons for amputation are continually decreasing. Trauma is responsible for 16% and cancer and congenital reasons merely 1% of all amputations [12,113]. Overall, dysvascularity is the most important and significant reason for amputation.

1.3 Level of amputation

Lower limb amputations can be performed at different levels according to the quality of the biological tissue. Amputation can be achieved by removing the toe, the foot, amputating through the shank (transtibial), the knee (knee disarticulation), the thigh (transfemoral) or by complete removal of the femur and femoral head (hip disarticulation).

In Canada and around the world, the most important level of lower limb amputation is the transtibial level (up to 65%), followed by the foot and transfemoral amputations [16,24]. The level of amputation is directly linked to sensorial and mechanical deficits. As well, the higher the level of amputation, the greater is the physiological demand [25]. Preserving as much biological tissue as possible provides important mechanical and physiological advantages for the transtibial amputee (TTA) when compared to the transfemoral amputee [26,27]. Therefore, when evaluating the level of amputation, health care professionals consider the level of amputation, which will provide greatest function as well as best health outcomes.

Though advances and improvements in technology, medicine, diagnosis and care have allowed for a lower level amputation (transtibial vs. transfemoral), important bilateral biomechanical differences are experienced by the TTA. The loss of foot and ankle joint function must be compensated for by other strategies (i.e. intact limb, intact joints of prosthetic limb). Thus, this poses additional demand on the intact limb and joints [28].

As well, the results of several studies have indicated that dysvascular and traumatic TTA should be considered as different entities when conducting gait analysis and establishing a rehabilitation plan [27,29,30].

1.4 Particularities of lower limb amputees

The largest TTA population is that of the dysvascular TTA (DTTA). The DTTA have related and concurring conditions or disease and therefore, to some degree, also have dysvascularity in the non-amputated limb [31]. Concurring diseases such as diabetic retinopathy, which affects vision, may also have an impact on gait. Indeed, a decreased capability of the visual system, which provides the CNS with primary afferent information with regards to postural stability, leads to important balance issues [32]. As stated above, this as well poses important demands to stability and gait and posture may therefore be impacted, even in the intact limb. Other concurring risk factors in DTTA are obesity, hypertension and overall reduced physiological function often leading to cardiovascular health risks [29,33].

Being sedentary and overweight/obese are important precursor to type II diabetes. It has been shown that often these two factors are part of a cycle, associated with other important and aggravating health conditions (i.e. cardiovascular disease), in the individual with type II diabetes [33]. Motivation and other like psychological factors play an important role in this vicious cycle, and if walking is to even be envisioned after amputation, these concerns must be addressed.

Indeed, three months after the amputation, only 61% of TTA are reported to be ambulatory and this number decreases to 51% 2 years post-operative, while others never regain full function and necessitate a walking aid [34,35]. As well, two systematic reviews concluded that DTTA have a significantly reduced ambulation rate when compared to their traumatic TTA counterparts though no statistics or numbers were provided other than the reduced VO_{2max} values observed in the DTTA (i.e. 26-29% reduced VO_{2max} in DTTA when compared to traumatic TTA) [36,37]. Age is another factor negatively correlated with ambulation and only about 2% of DTTA older than 85 years old are able to relearn walking [38,39]. Also, without change to hygiene and management of vascular issues (i.e. diabetes management, foot care, etc.), amputation of the contralateral limb is commonplace and further mechanical and physiological limitations are implicated with this second amputation [40]. These factors combined place the DTTA at greater risk of loss of autonomy.

Survival rate and overall life expectancy are lower in the DTTA population when compared to traumatic TTA [30]. Of those who suffer an amputation for vascular reasons, 50% will die within 5 years [114]. Also, over 15% will suffer a contralateral limb amputation within the next year, this number doubling for every year following the amputation, which causes further reduction in function and increases physical demands and biomechanical deficits [41]. Therefore, though improving life expectancy and reducing morbidity risk are essential, improving the quality of life in the DTTA is also of importance. Preserving walking in the DTTA is imperative to maintain their autonomy.

Additionally, the prosthesis with which is equipped the TTA takes habituation for posture and locomotion as well as pain management. Fitting is very much trial and error, and often, several adjustments must take place before the best fit can be provided to the TTA. A thorough analysis of prosthetic devices is beyond the scope of this thesis, but the main two categories of prosthetic devices are presented here: passive and powered prosthetics. Passive prostheses are non-motorized, simple foot and rod mimicking the likes of the missing human foot and shank. These prostheses often have different composite materials aiding in providing some joint movement at the ankle as well as cosmetic material covering. These materials allow some compliance when body weight is put on the prosthetic, compliance, which is then restored to the TTA during toe-off [122,125]. Still other passive prosthetics called energy-storage-and-return, are designed to further allow for compliance and return of this stored energy to enable toe-off [41]. On the other hand, powered prostheses are designed to mimic the various missing anatomical structures, which provide propulsion to walking [147]. Powered prosthetics research is an area of increasing interest, but many obstacles are ever present in the design of these. Mimicking intact limb structures and power output has been shown to cause falls, and thus, powered prosthetic designers must build prosthetics that propulse during walking, without compromising stability [148].

Unfortunately, less than 43% of the DTTA patients have completed an inpatient rehabilitation program though these protocols have been associated with improvement in survival rate and general health as well as a decreased risk of re-amputation [115]. The following sections will explore walking in the TTA and DTTA when compared to healthy adults, in understanding the specific biomechanics of each population throughout gait. Then, in starting from a position of quiet standing, the initiation of gait will be compared between the DTTA and their healthy counterparts.

2. Steady-State Walking

2.1 Gait in Healthy Adults

Walking is the most common form of human locomotion and it is involved in almost all activities of daily living [28]. As well, walking is a fundamental building block to many more complex movements such as running and other sport and daily movements [79]. The primary goal of walking is to propel the body forward while maintaining posture and balance, all the while resisting gravity [80]. In order to accomplish this, walking is controlled by numerous muscles activated in sequence.

When born, it takes a child approximately 8 months (i.e. 6-12 months) to learn to stand erect using support [81,82,116]. However, on average, an additional 4 months are needed for acquiring autonomous walking. Walking is continually refined up until the age of approximately five years old, as children learn to narrow their stance, activate muscles in a synchronous and efficient manner, increase knee flexion during swing, etc. [83]. After the age of five, changes observed in gait are due to change in size and stature as children grow until 12 to 18 years old, depending on sex and genetics, and thus, gait changes are minor in adolescence [84]. Further differences are only seen in the presence of important morphological modification, pathology and/or advanced age [85].

Likewise, the absence of walking or gait refinement in children as they age is an important indicator of disorder. Many indices and other like gait analysis tests have been formulated to assess for these developmental delays [86]. Then, in elder adults, important changes in walking are seen starting approximately at the age of 65 years of age [10,87,88]. Health and physical fitness level play an important role with regards to these changes in walking. Again, presence of pathology in the elderly most often impacting the walking pattern.

Because walking is an important indicator of mobility, it also defines independence. Thus, in the presence of pathology and older age, the ability to walk is an important indicator of overall health [1]. Because of the wide variety of disorders and pathologies, the impact on gait takes place in various ways and forms. To understand how pathology and disorder impact gait, gait analysis is therefore essential.

Muybridge and Marey were the first scientists to document on the mechanics of gait in the late 1800's [2]. Using a series of photographs, Muybridge and Marey captured the various movements and phases involved in various human and animal movements. The images captured by the photographs provided an important basis for today's gait analysis. This was the first attempt into understanding the mechanics of the complex task of walking [89]. Thereafter, during World War II, there was an increase in the numbers of amputations and with the need to build appropriate prostheses, gait biomechanics was, once again, the area of great interest. Finally, in the 1980's, biomechanics came into study as a new field of research. Hence, the number of studies, pertaining not only to biomechanics but also with regards to gait, importantly boomed. For example, a simple PubMed search with the keywords *human gait* reaches over 41 000 publications (January, 2018).

From this early period in biomechanics, the techniques, the measurement instruments, their precision and the knowledge concerning gait analysis have been developed in overwhelming quantity and variety. Today, in an era of evidence-based medicine, the need for quantitative analysis is imperative and important advancements in this regard have been made. Indeed, such tools as sophisticated motion capture and analysis systems are able to record and provide quantitative kinematic gait information with high precision and three-dimensional (3D) reconstruction of movement and segment modeling are possible. Force platforms, recording CoP and ground reaction forces (GRF), provide precise and minute kinetic data of gait, allowing to understand not only the movement created about various segments and joints, but rather how and through which means this movement is created (CoP and GRF are described below in sections 3.1 and 4.1, respectively). By quantifying the biomechanics of human movement, it is possible to observe differences intra-individually. Comparisons can also be made inter-individually to compare pathology with healthy controls. To quantify these differences and the walking pattern, spatiotemporal, kinematic and kinetic parameters have been investigated.

The gait cycle has been extensively studied and outlined and various terms have been used to describe it. The current thesis will describe that provided by Whittle and is illustrated below in Figure 1 [90]. One complete gait cycle is defined from the heelstrike of the right or left foot to the subsequent heelstrike of the same foot. During this gait cycle, the limb goes through a support and swing phase. During the support phase, the foot is in contact with the ground, from heelstrike through toe-off. During the swing phase, the foot leaves the ground and the leg swings forward, with the knee flexed, and ends at the next heelstrike. In healthy adults, support phase is approximately 60% of the gait cycle and swing phase accounts for approximately 40% of the gait cycle. The support phase can be divided into a single support phase (40% of gait cycle), when only one limb is in contact with the ground, and two double support phases (each of 10% of the gait cycle) when both feet are in contact with the ground [90].

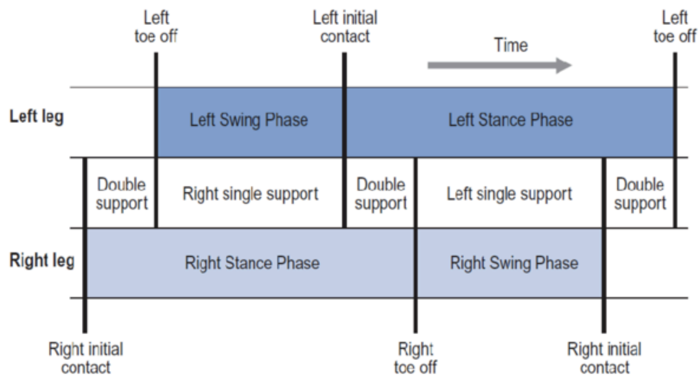


Figure 1- Illustration of gait cycle phases. One complete gait cycle of both the right and left limbs is illustrated with respect to time (Adapted from Whittle (1996) [90]).

Within each support and swing phases, the limb goes through various sub-phases. Again, the following sub-phases of the gait cycle described below are an excerpt from Whittle (1996). As soon as the first heelstrike takes place, both feet are simultaneously in contact with the ground (i.e. first double support period). There is then a loading response in which the knee flexes, in order to cope with the large amount of body weight that is placed on the limb. At this point, the lowest height of the center of mass (CoM) is reached. Next, during midstance, the knee extends to a straight leg as the body travels over the standing limb. The maximum height of the CoM is reached at full knee extension. During terminal stance, the heel rises from the ground as the opposite heelstrike occurs, initiating the second double-support period. The ankle plantarflexors are actively involved in pushing the limb into swing, creating the second single limb support phase. Finally, during swing, the knee flexes in order to help in swinging the limb forward, and the knee then extends for the next heelstrike. As this subsequent heelstrike occurs, the gait cycle is repeated once again. The gait cycle described above is also known as the stride cycle, when referring to each limb.

Spatiotemporal parameters are those parameters which quantify both time and distance of the different phases during gait. As such, walking velocity is the defining parameter for steady-state gait [91]. As stated, average SSWV is 1.3 to 1.5 m/s in healthy adults, and this SSWV has been shown to be dependent on stature, weight and other non-anthropometric or morphological characteristics (i.e. maximum oxygen consumption) [57,92]. When walking at SSWV, studies have shown that energy efficiency is optimized for an individual, as walking at a slightly reduced or faster walking velocity will result in increased metabolic demands [93]. Other common spatiotemporal parameters include, and are not limited to, cadence, stride/step length, stride/step width, stride/step time and single/double support and swing times.

Interestingly, most spatiotemporal parameters are related to walking velocity. By increasing either cadence or stride length for example, walking velocity can be augmented. As well, stride length is a limiting factor to increasing walking velocity as it is a function of stature and functional range of motion (i.e. length of limb, flexibility, etc.) [94]. Because walking velocity is an encompassing variable, its measurement is significant and allows insight into the other parameters which it modulates.

Numerous studies have outlined a reduction in SSWV in older adults, with the onset of this reduction appearing around 55 to 65 years old, dependent on physical fitness levels, health and other such factors [93,95]. Additionally, there is a significantly greater reduction of SSWV in the elderly affected by different pathologies [78,96].

Along with the spatiotemporal parameters which describe gait, the joint kinematic parameters during SSWV follow a specific pattern in healthy adults. The average sagittal plane joint angular kinematics of the ankle, knee and hip are outlined below in Figure 2 in 19 healthy adults walking at SSWV and will be discussed in the following paragraphs with respect to SSWV in healthy adults [10].

At heelstrike, there is hip flexion of the amplitude of about 20°. Then, as the body travels over the leg to final push-off, the hip progresses from flexion to a maximum extension of approximately 12°. Finally, at heel-off through to swing, there is a flexion of the hip, maximum flexion (i.e. approximately 20°) observed during late swing.

The foot contacts the ground with the knee fully extended. Then, with weight acceptance, the knee flexes to about 20°. As the body travels over the foot during midstance, there is then an extension of the knee to about 5° of flexion, and this is as well associated with the lightening phase discussed further below with regards to the vertical GRF (i.e. minima value in midstance). Finally, with final push-off and swing, the knee flexes to a maximum of about 60° reducing its radius of gyration to allow the leg to successfully swing forward, then extending at the end of swing for the next heelstrike.

Finally, with regards to the ankle, there is a small plantarflexion of about 3° after initial heelstrike to flatfoot. With the phases of weight acceptance and midstance, there is progressive dorsiflexion to a maximum of about 10°. Perhaps the most important joint movement, as well discussed further below with regards to moment and power, comes with the maximum ankle plantarflexion amplitude of approximately 20°, at the push-off phase just prior to toe-off. There is then rapid dorsiflexion during early and midswing to help foot clearance. Then, the ankle is kept approximately in neutral position (i.e. 0° plantarflexion/ dorsiflexion) until the next heelstrike.

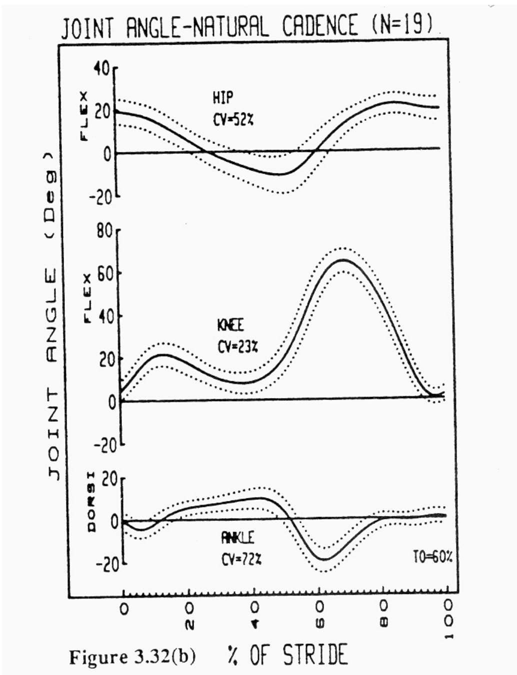


Figure 2- Joint angular kinematics of the lower limb. Average (solid line) and coefficient of deviation (dotted lines) joint angles of the ankle, knee and hip in 19 healthy adults for one stride cycle. Positive values display flexion, flexion and dorsiflexion at the hip, knee and ankle, respectively (Adapted from Winter, 1991 [10]).

Though the movement of the lower limbs (i.e. kinematics) during the gait cycle has been discussed, with the accompanying spatiotemporal parameters, the underlying forces which produce these movements are also of importance. Thus, the interaction of the CoP and CoM as well as the underlying kinetics during SSWV in healthy adults will be discussed in the following paragraphs.

Walking is a unique example of the regulation by the CoM by the CoP. When walking, the CoM is projected beyond its base of support and caught again within the base of support with the heelstrike of the next step. This pattern is repeated as the CoM is pushed along, being caught and pushed again from step to subsequent step [97].

At a SSWV, the vertical GRF pattern beneath each foot is that of double hills with a valley. The first peak represents initial loading of the limb after heel contact and the magnitude of the force rises to 11 N/kg (i.e. 1.1 times body weight (BW)). The slope of the vertical force from heel contact until maximum weight acceptance (first peak of vertical GRF pattern in Figure 8) is referred to as the loading rate (N/s). Hence, the steeper the vertical force time profile, the greater the loading rate [147]. The valley, between both vertical GRF peaks, represents a lightening phase during midstance (described above) with a force of approximately 7.5 N/kg (i.e. 0.75 times BW). This is caused by the ipsilateral knee extension and the contralateral toe-off and knee flexion, which aids in moving the leg upward into swing. The second peak represents push-off from the plantarflexors through to toe-off, the magnitude of which is approximately 11 N/kg (i.e. 1.1 times BW). This vertical GRF pattern is outlined in Figure 3 below.

During SSWV, the typical GRF pattern in the antero-posterior (A/P) direction is an initial braking force followed by a propulsive force. The braking GRF are represented as negative values and the propulsive GRF are represented as positive values in Figure

3. At SSWV, the braking and propulsion GRF are equal. If the force applied in the anterior and posterior directions are not equal, there will be an increase or decrease in walking velocity, depending on whether braking or propulsion force is greater [68]. In SSWV, the peak magnitude of both the braking and propulsive forces are approximately 2 N/kg (i.e. 0.2 times BW). The midpoint, at which the braking force switches to a propulsive force, is at 50% of the stance phase, or 30% of the total stride cycle, as outlined in Figure 8.

The medio-lateral (M/L) GRF applied during SSWV are omitted from the current thesis as they did not contribute to forward propulsion of walking.

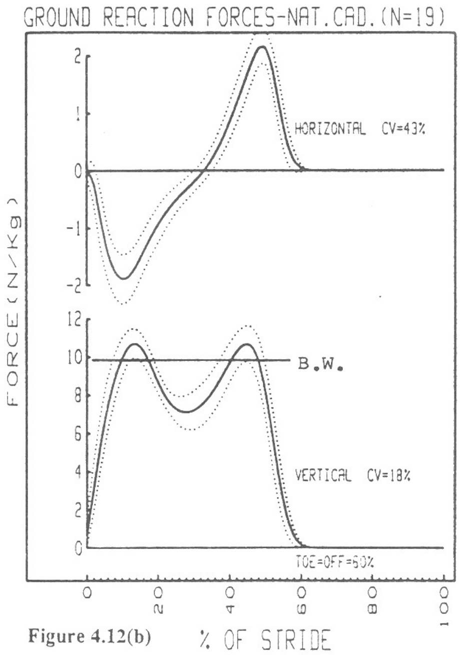


Figure 3- A/P and vertical GRF profiles during SSWV (top to bottom). Average GRF pattern in 19 healthy adults during walking at SSWV expressed in N/kg and over 100% stride cycle. Positive values describe propulsive A/P and vertical GRF. BW is outlined by the straight line B.W. on the vertical GRF profile (Adapted from Winter, 1991 [10]).

Moment and power are also kinetic parameters, which are calculated through the combination of anthropometrical data, kinematic and kinetic parameters with an inverse dynamic approach [28,66,98]. Thus, the net moment at the ankle, knee and hip can be estimated. Moments at the joint provide insight into the net muscle effect on the joint. For the current analysis, moment at the ankle, knee and hip will be considered positive moments if they produce net moment about the joint in resistance to gravity and forward motion. That is, the extensor moment produced at all three lower limb joints. The average sagittal plane moment at the ankle, knee and hip joints and total support moment for 19 healthy adults is outlined in Figure 4 and are discussed hereafter with regards to the net muscle moment active about each lower limb joint [10].

A small net dorsiflexor moment, created by the tibialis anterior contraction, is first present at the ankle during heelstrike to help in lowering the foot to the ground (i.e. from heel contact to foot flat in weight acceptance) [99]. Then, there is a plantarflexor moment created by the gastrocnemius- soleus muscle complex which starts after weight acceptance and grows to about 2.0 Nm/kg prior to push-off. This muscle moment is the most important propulsive contribution to walking. That is, approximately 80% of propulsion can be accounted for by the moment created at the ankle during push-off [10]. Thus, the absence of the ankle joint produces important deleterious effects to forward propulsion. The important contribution of this plantarflexor moment to walking is discussed further with regards to power.

At the knee, there is an extensor moment produced by the quadriceps muscles in early stance to control the knee flexion during the weight acceptance (i.e. about 0.5 Nm/kg). There is then a net knee flexor moment during midstance created by the hamstrings (i.e. about 0.2 Nm/kg). Just prior to and after toe-off, there is a small net knee extensor moment again produced by the quadriceps, which acts in controlling the knee flexion caused by the strong ankle push-off (i.e. about 0.1 Nm/kg). Just prior to next heel contact, during late swing, there is a net knee flexor moment by the hamstrings which helps in decelerating the leg and foot (i.e. about 0.15 Nm/kg).

At heelstrike, there is an initial net hip extensor moment (i.e. about 0.4 Nm/kg) by the gluteal muscles which aids in preventing the knee from collapsing at heelstrike and weight acceptance phases. This is followed by a net hip flexor moment created by the iliopsoas muscles, from mid-stance to early swing (i.e. maximum net moment of about 0.4 Nm/kg). Finally, during the latter half of the swing phase, the net extensor moment observed at the hip by the gluteal muscles helps in decelerating the thigh and leg as it swings forward for the subsequent heelstrike, with a small moment of about 0.3 Nm/kg.

As stated, positive moments describe the extensor moment at all three joints which resist gravity and produce forward progression. The summation of these three moment profiles is termed the support moment, defined by:

$$M_{support} = M_{ankle} + M_{knee} + M_{hip} \qquad \text{(Equation 1)} \qquad [99]$$

The magnitude of this support moment peaks at approximately 1.0 Nm/kg during weight acceptance and terminal stance. Finally, it has been noted that the pattern of support moment resembles that of the vertical GRF as the support moment is a clear reflection of the forces applied to the ground [98].

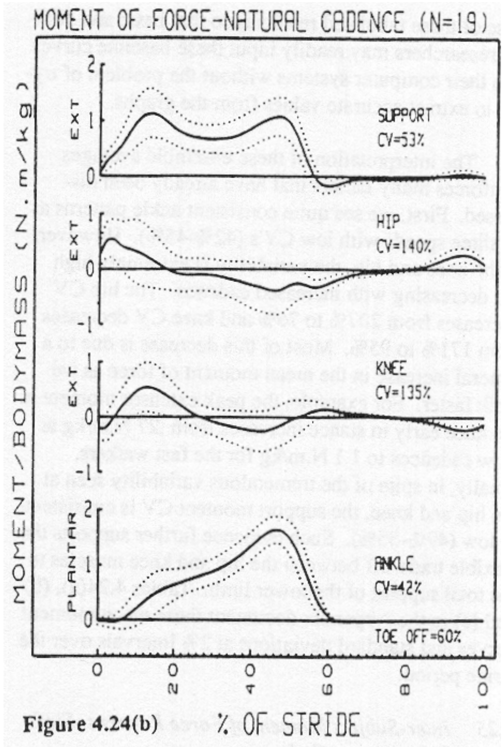


Figure 4- Ankle, knee, hip joint and support moments during SSWV (bottom to top, respectively). Positive moment is defined as extension for the support, hip and knee moment and as plantarflexion at the ankle joint. Average sagittal plane joint kinematics for 19 individuals walking at natural cadence (Adapted from Winter, 1991 [10]).

Power is an additional parameter which can be obtained by combining kinematic and kinetic data [28]. Power (in Watts) at a given joint (P_j) is obtained by combining the net joint moment (M_j) with angular joint velocity (ω_j).

$$P_j = M_j \times \omega_j \qquad \text{(Equation 2)}$$

When the moment and angular joint velocity have the same polarity, this is defined as power generation created by a concentric contraction of the muscle group involve. Power absorption is defined as when the joint moment and angular joint velocity have opposite polarities and the muscle group is acting in an eccentric contraction [28].

Figure 5 below outlines the average sagittal plane power profiles at the ankle, knee and hip during SSWV in 19 healthy adults. The various peaks throughout the respective power profiles have been identified at the ankle (A1, A2), the knee (K1, K2, K3, K4) and finally at the hip (H1, H2, H3). These specific bursts will be discussed below with regards to each joint for the data displayed in Figure 5 [10].

At the hip, the three bursts of power can be defined by two power generation (H1 and H3) bursts and one power absorption burst (H2). H1, present after heelstrike and throughout weight acceptance, is marked by power a generation created by the extending hip angle and the net extensor moment about the hip (i.e. about 0.2 W/kg). Then, during midstance to late stance (20-50% of stride), there is the H2 power absorption burst created by the extending hip angle and the net hip flexor moment. The H2 peak power absorption burst has been documented at about -0.2 W/kg. Finally, during push-off through to initial swing, there is a final H3 power generation burst of approximately 0.6 W/kg peak, the most important power burst at the hip in terms of magnitude, associated with the action of the flexing hip and the net flexor moment created at the hip.

At the knee, the K1 burst is the first power absorption during weight acceptance as the knee is flexing and a knee extensor moment is present (i.e. approximately of -0.5W/kg). Then, the K2 burst (0.2W/kg) occurs during midstance as the knee is extending and an extensor moment is present producing power generation. This is the sole power generation produced by the knee. From just prior to toe-off through mid-swing, as the knee is flexing under a net knee extensor moment, there is again a power absorption burst with a peak of about -0.5W/kg, identified as K3. Finally, during the latter half of the swing, the knee joint starts to extend as the knee moment also reverses to a flexor moment, thus creating the last K4 power absorption burst of approximately -0.5 W/kg. Thus, the overall power output of the knee is absorption, rather than generation [28].

After initial heelstrike there is a net dorsiflexion moment as the ankle is plantarflexing, causing the first ankle absorption burst, A1. A1 continues as the ankle is dorsiflexing in weight acceptance and a net plantarflexion moment is created about the ankle. The maximum magnitude of A1 is approximately -0.5 W/kg. The most important power generation is then produced at the A2 power burst, present from about 40 to 60% of the stride cycle. The large A2 power burst coincides with the second peak of the vertical GRF and the peak propulsive A/P GRF, making A2 the primordial contributor to biomechanical energy in walking [28]. This A2 power generation peak is of the magnitude of approximately 3.5 W/kg and coincides with the maximum plantarflexor moment as the ankle is plantarflexing, created by the gastrocnemius-soleus contraction and the important amplitude of ankle plantarflexion movement displayed at the ankle producing an important push-off.

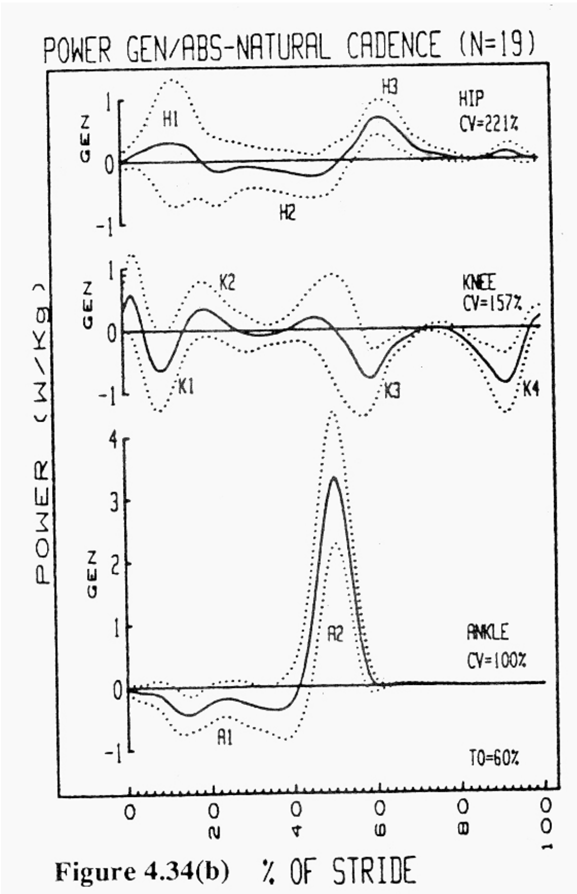


Figure 5- Power profiles of the ankle, knee and hip joints (from bottom to top, respectively). The average sagittal plane power profiles of 19 individuals during SSWV are expressed in W/kg (Adapted from Winter, 1991 [10]).

Thus, in the absence of plantarflexor muscles (i.e. gastrocnemius-soleus muscle complex) and with the amplitude reduction of ankle range of motion (i.e. prosthetic foot), gait is inevitably impacted. The following section will therefore characterize the gait pattern in such a situation, that of the TTA and DTTA.

2.2 Gait in the TTA and DTTA

As will be seen with quiet standing and gait initiation, important biomechanical differences are present in the TTA and DTTA when compared to healthy adults during SSWV. Moreover, important differences are also denoted between the traumatic TTA and DTTA. These differences will be compared with regards to spatiotemporal, kinematic and kinetic parameters. Finally, the relevance of these parameters for gait analysis will be discussed.

First, with regards to spatiotemporal parameters, it is known that the TTA walk at a reduced SSWV when compared to able-bodied individuals. This velocity has been shown to be approximately 1.0 m/s [77] when compared to about 1.3 to 1.5 m/s in healthy adults [118,119,120,121].

The stride length and cadence during SSWV in the TTA are also reduced when compared to controls [109,117]. This is to be expected as both stride length and cadence contribute to the calculation of walking velocity (i.e. Equation 1).

Important differences are also present between the intact and prosthetic limb throughout SSWV in the TTA [109,129]. There is an increased contribution of the intact limb to forward motion (i.e. examined below with regards to GRF), as there is also increased loading on the intact limb when compared to the prosthetic limb throughout gait [144]. As well, TTA spend less time on their prosthetic limb when compared to their intact limb when walking. Total stance time is increased, especially in the intact limb, and double support time is also increased [145,146].

As with gait initiation, the most important change to kinematics during SSWV in the TTA is due to the missing ankle joint. That is, when examining the sagittal plane ankle angle kinematics, the 20° plantarflexion angle observed during push-off in healthy adults is approximately 5° in the prosthetic limb [122].

In studies observing kinetic gait differences in TTA, it was found that TTA exhibit a decreased knee moment and increased hip moment contribution at SSWV. In addition, TTA exhibit a passive moment at the prosthetic ankle [11,123,124,125]. Figure 6 below outlines the average support, hip, knee and ankle moment profile in eight below-knee amputees (adapted from Winter & Sienko, 1988 [11]).

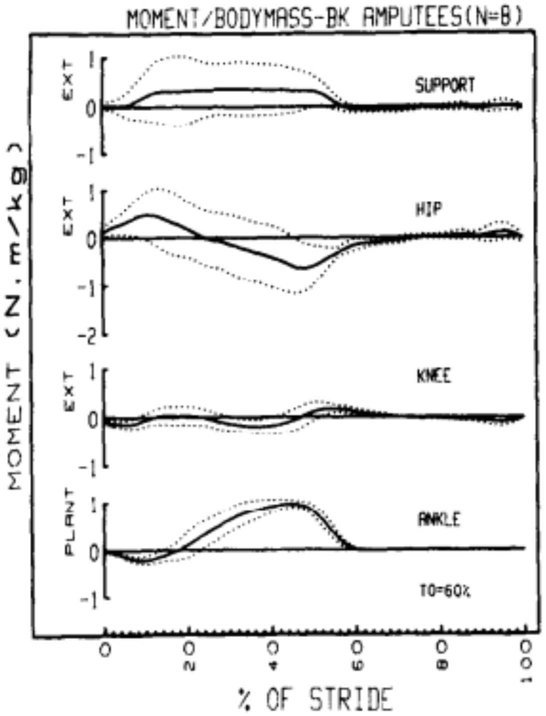


Figure 6- Average sagittal plane ankle, knee, hip joint and support moments in 8 below-knee amputees (Top to bottom, respectively). Moments are expressed as a percentage of total stride cycle and in Nm/kg (Adapted from Winter & Sienko, 1988 [11]).

The most important changes to gait in the TTA are due to the loss of the ankle plantarflexors, which permit propulsion, whole body support and initiation of limb swing [6,11,126,127]. The prosthetic foot provides a passive ankle moment, as the force and moment created at the prosthetic ankle are generated by the BW applied to the prosthesis (i.e. compliance). This compliance of the prosthesis has been shown to provide some energy storage and return as it bends under the BW and energy is returned as BW is unloaded. That is, as the ankle dorsiflexes during early stance phase and through to flat foot, energy is stored to the prosthesis which is then released during late stance push-off phase. However, this mechanical energy released by the prosthesis has been documented to be less than half release of that usually produced by the ankle plantarflexor muscles [100,122,128]. Most studies investigating the compliance and energy return of the prosthetic feet have done so in comparing various types of prosthetics. The comparison of various prosthetic devices is, however, beyond the scope of this thesis.

Finally, due to the loss of ankle joint at the prosthetic limb, there are compensatory biomechanics produced. As observed in Figure 7 below, a reduced A2 power generation phase is accounted for by the loss of the plantarflexors in the prosthetic limb (i.e. about 0.2 W/kg vs. 3.0 W/kg in TTA vs. controls, respectively). At H1 in Figure 7, there is considerably more power generated by the hip extensors when compared to controls (i.e. about 0.5 W/kg vs. 0.2 W/kg in TTA vs. controls, respectively). Winter and Sienko (1988) have proposed that the extensor gluteus maximus muscle of the hip somewhat compensates for the missing plantarflexors, aiding the below-knee amputee to propel the body forward [11]. Figure 7 below outlines the average sagittal plane power profiles in 8 below-knee amputees for the hip, knee and ankle joints.

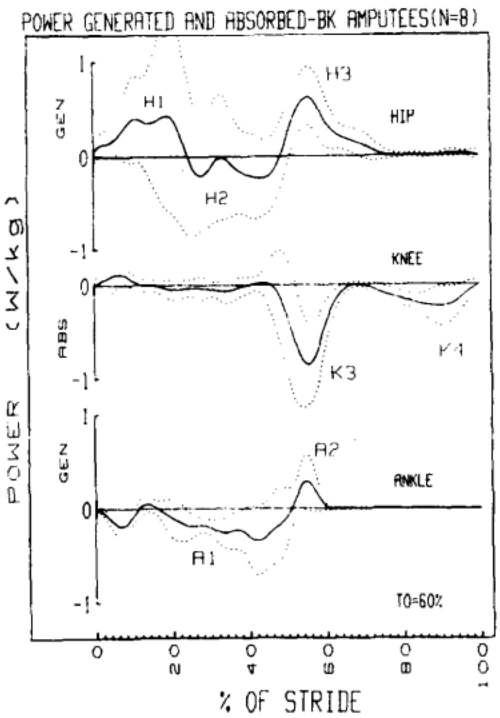


Figure 7- Average sagittal plane ankle, knee and hip power profiles in 8 below-knee amputees (top to bottom, respectively). Expressed as a function of stride cycle and in W/kg (Adapted from Winter & Sienko, 1988 [11]).

The many spatiotemporal, kinematic and kinetic parameters presented in the above section on gait at SSWV as well as will be presented in the below sections on quiet standing and gait initiation, make proof of the wide array of parameters available for gait analysis, both in healthy adults and amputee populations. Thus, when selecting parameters to observe during gait analysis, scientific rigor is necessary in establishing the best parameters with which to conduct gait analysis. The following paragraphs will therefore explore the topic of parameter relevance for gait analysis.

The differences denoted in the TTA and DTTA testify of the importance of gait analysis as a mean to diagnose pathology, set a prognosis and establish and evaluate a treatment plan [101,132]. A wide diversity of parameters of various types exists and are readily used to examine and explain human gait [128,130,131,134]. Also, though the variety of equipment available today in clinical settings, gait analysis is often carried out solely through clinician observation [102]. Although clinicians have developed good expertise through many years of practice and training, clinician observations (i.e. most often with regards to angular kinematics and differences between the limbs) remain subjective [103,133]. Principal reason for main, and perhaps sole use of clinician observation as means of gait analysis, is ease of measurement and cost efficiency [131,133,135].

Motion capture and analysis systems, force transducers and transmitters are but examples of the wide variety of tools which exist and have been developed for gait analysis. Additionally, a wide selection of specific products and brands exist within each category of gait analysis tools. Therefore, the variety and amount of quantitative data possible is seemingly infinite and selecting which data, gait analysis method and tools for data collection can be a challenge in itself. Indeed, selection of method, measures and tools of measurement for gait analysis is of primordial interest. Moreover, before and after data collection, decision with regards to appropriate gait parameter for gait analysis is even more important.

In accordance with evidence-based-medicine, the biomechanical parameters chosen are important to rigorous gait analysis [104]. Because of the large number of parameters available, it seems reasonable that certain parameters would be best suited for gait analysis in specific populations.

Thus, systematic reviews are realized in an attempt to organize and add understanding to the practice of gait analysis in various populations. Sagawa and colleagues conducted a systematic review of all applicable studies, to determine the most relevant biomechanical and physiological parameters for gait analysis in the lower limb amputees. The term relevant was defined as those parameters commonly used, able to discriminate and/or have specific clinical relevance for the gait analysis of lower-limb amputees. These studies were selected based on inclusion and exclusion criteria, constituting a review of homogeneous studies with regards to lower-limb amputee gait analysis. In this unique systematic review, by pooling such studies and tabulating the measured biomechanical parameters, a certain valuation of the biomechanical parameters for lower-limb amputees was carried out. Physiological and other parameters explored by Sagawa et al. are beyond the scope of this study and will thus be omitted from this current thesis.

As well, a Level of Evidence score (out of 13) was given to each of the 89 articles included in the Sagawa systematic review based on: 1) selection of participants, 2) intervention and assessment, and 3) statistical validity. The Level of Evidence score was then compared to the Impact Factor of the journal in which each included article was published.

No relation was found between the established Level of Evidence and the Impact Factor. As well, most articles (i.e. 73%) did not have sufficient participants for statistical validity and thus, in interpreting the results, Sagawa and colleagues recommended the reader to be cautionary. The authors concluded that a wide variety of parameters were pertinent to gait analysis in lower-limb amputees, namely walking velocity, and other like spatiotemporal parameters (i.e. cadence, step length and stride length), joint angular position of the lower limb joints and kinetics recorded from force platforms below the feet providing GRF and impulse parameters. However, the authors concluded that there was a lack of consensus among the included studies, and other systematic reviews should be carried out in the hopes of providing a more evidence-based approach to gait analysis [105]. The measured parameters in the 89 included articles are outlined below in Table 1.

Biomechanical parameters			**Physiological parameters**	**Other parameters**
Spatio-Temporal 153	**A. angles (deg) 78**	**Platforms 72**	**Cardiologic 9**	
Speed (m/s) 43	Trunk angles 5	GRF transverse (N/kg) 1	Heart rate (bpm) 8	RPE 4
Cadence (step/min) 16	Pelvis angles 8	GRF anteroposterior (N/kg) 14	Blood pressure (mmHg) 1	Satisfaction 5
Stride t. (s) 4	Hip angles 12	GRF vertical (N/kg) 30		Comfort 1
Stance t. (s) 16	Knee angles 31	GRI anteroposterior (N.s/kg) 17	**Respiratory 45**	Preference
Swing t. (s) 5	Ankle angles 22	GRI vertical (N.s/kg) 3		Comorbidity n. 1
Step t. (s) 3		GR moment (N.m/kg) 1	VO_2 (ml/min/kg) 30	Cost components 3
Single sup. t. (s) 2	**A. moments (N.m/kg) 58**	COP (m) 6	VO_2 cost (ml/m/kg) 13	Falls n. 3
Double sup. t. (s) 3			Respiration rate (bpm) 1	Osteoarthritis clinical diag. 2
Foot flat t. (s) 3	Hip moment 13	**Accelerometer (m/s²) 1**	Respiratory quotient (%) 1	Socket displacement 2
Stance t. ratio (%) 3	Knee moment 27			Socket volume 2
Swing t. ratio (%) 2	Ankle moment 18	Acc vertical 1	**EMG 26**	
Step length ratio (%) 9				
Timing (%) 1	**A. powers (W/kg) 64**	**Pressure 8**	EMG lower limb muscle intensity 9	
Stride length (m) 17			EMG lower limb muscle time 17	
Step length (m) 20	Trunk power 1	Pressure socket time 1		
Step width (m) 1	Hip power 26	Pressure socket intensity 7		
CV (%) 3	Knee power 21			
Steps week 1	Ankle power 16	**M. dynamometer (N) 1**		
Timed walk test 1				
	C. of Body mass (m) 6	Manual hip force 1		

Table 1- Frequency of parameters measured in the 89 articles included in the systematic review of lower limb amputee gait by Sagawa et al. 2011. Type of parameters (frequency of measurement) are listed as headings and individual parameters (frequency of measurement) of same type are listed beneath (Adapted from Sagawa et al. 2011 [105]).

Sagawa and colleagues' approach in tackling the tough question of parameter relevance included a very broad population, that of lower limb amputees. Knowing that the unilateral and bilateral, the transtibial and transfemoral and finally, the traumatic and dysvascular amputee pose very specific constraints in gait, the most relevant parameters for gait analysis are perhaps varied throughout these numerous specific populations (i.e. DTTA vs. traumatic bilateral transfemoral amputee).

In the clinical setting, gait analysis must be cost efficient and ease of measurement is also important. The numerous parameters identified by Sagawa et al. are perhaps not time and cost efficient in the clinical setting. However, the various spatiotemporal parameters, such as walking velocity, are cost efficient and can be measured with ease.

The results obtained by Sagawa and colleagues leads to question whether the same biomechanical parameters are most relevant for gait analysis in healthy adults. Indeed, knowing the most relevant parameters for gait analysis in healthy adults would allow for comparison with pathologic populations and serve as reference for gait analysis. Thus, a systematic review evaluating the most relevant parameters for gait analysis in healthy adults is warranted to compare with the lower limb amputee population.

3. Quiet Standing

3.1 Quiet Standing in Healthy Adults

When standing upright, and seemingly immobile, an individual is constantly making minor adjustments in response to external and internal stimuli. Thus, the term quiet standing is used to describe this state [8].

In healthy adults, quiet standing has been explained via the theory of the inverted pendulum [42]. Two important variables with regards to biomechanics are here discussed, first introduced in section of SSWV above: total body CoM and net center of pressure (CoP_{net}). CoM is a point at which the total body mass can be averaged. The CoM is a passive variable whose position is expressed in metres within the 3D space [43]. The CoP_{net} is a point which represents the position of application of the resultant forces under the foot/feet during standing and walking. The CoP_{net} can be calculated from the orthogonal forces and moments recorded from a force platform. Individual right and left CoP can be measured beneath each foot with two separate force platforms. The CoP_{net} is the weighted average of both the CoP_{right} and CoP_{left} [43].

In an effort to keep the CoM within the base of support, the CoP_{net} is continually moving in such a way as to catch up and regulate the CoM (i.e. maintain balance). That is, the CoP_{net} is the independent variable which modulates the CoM position. In quiet standing, the goal is to keep the CoM within the base of support. If the CoM moves beyond the base of support, important action must take place to avoid falling. Most often a step is taken in the direction of the excursion of the CoM to catch the CoM and bring it back within the new base of support [8]. Such calculations as the CoP-CoM parameter, which represents the distance between the CoP and the CoM in terms of the root mean square (RMS), have been used to reflect postural sway, and thus postural control [149].

Control of the inverted pendulum is made possible by the visual, vestibular and somatosensory systems. Thus, if one or more of these systems is absent or impaired (i.e. eyes closed or sensory loss with amputation), control of the inverted pendulum is challenged [44]. Signs of reduced balance control is associated with an increased risk of fall and postural sway provides an objective measure of balance control [45,46,47]. Postural sway has been defined as the A/P and M/L amplitudes of the CoP_{net} displacements during quiet standing [8].

3.2 Quiet Standing in the Transtibial Amputee

In a systematic review of quiet standing studies conducted among lower limb amputees, it was concluded that greater imbalance is observed in this population when compared to healthy adults. Postural stability was observed as a function of the sway amplitude, sway velocity, total sway area as well as RMS amplitude of CoP_{net} in both A/P and M/L directions [48]. The A/P and M/L CoP_{net} amplitude is defined by the distance between the maximum excursion points of the CoP_{net} in opposite directions (i.e. measured in mm). A/P and M/L CoP_{net} velocity is defined as the distance travelled by the CoP_{net} over time (i.e. measured in mm/s). The CoP_{net} RMS is used to quantify the deviations from the mean of CoP_{net} velocity or amplitude displacement. The overall increase in the amplitude and velocity of CoP_{net} observed in the review of TTA posture by Ku et al. can be explained by asymmetrical and greater loading placed onto the intact limb [48,49,50,51].

Balance has also been investigated in the DTTA and traumatic TTA. Hermodsson and colleagues (1994) assessed quiet standing and one limb stance in the DTTA, the TTA and healthy adults. When quiet standing was measured, the DTTA demonstrated increased sway in the M/L direction when compared to the traumatic TTA and control subjects (i.e. measured as the standard deviation of the M/L CoP_{net} amplitude). Interestingly, no increased A/P sway was observed in the DTTA when compared to controls, and the traumatic TTA group demonstrated significantly decreased A/P sway when compared to controls. It has been theorized that this may be due to the stiff ankle created by the prosthesis in all TTA, as a prior study had suggested [52]. Thus, the traumatic TTA would appear to better maintain control in the A/P CoP_{net} direction as well. Hermodsson and colleagues conclude that postural stability measures discriminate between the DTTA and traumatic TTA [29]. Such a study warrants the need to differentiate between cause of amputation when conducting postural analysis, as reason for amputation creates unique underlying biomechanics [48].

In quiet standing, the CoM projection is maintained within the base of support by the movement of the CoP_{net}. Locomotion requires unbalancing of the CoM to produce forward progression [8]. The following sections will therefore address the biomechanics of gait initiation in healthy adults, TTA and DTTA populations.

4. Gait Initiation

4.1 Gait initiation in Healthy Adults

Along with quiet standing, gait initiation is involved in each walking bout. Gait initiation has been defined as the transitory state from quiet standing to steady-state walking [53]. The goal in gait initiation is to go from a state of static balance, with the CoM safely within the boundaries of the base of support, to a state of dynamic balance (i.e. controlled imbalance) [54]. For the purpose of the current thesis, gait initiation will be analyzed from the APA prior to gait initiation through to the forces produced by the first step to accelerate the CoM and until the SSWV is reached. As well, for the current analysis, the limb which initiates the first step will be termed the leading limb. The trailing limb will define the limb which is loaded as the leading limb initiates the first step.

Gait initiation can be described with various biomechanical variables such as the CoP_{net} movements beneath the feet in both A/P and M/L directions, known as the APA's, the kinematics of segments and the kinetics exerted under the activity of muscle contraction. This section will first explore the APA's which precede, and take place during, gait initiation.

A particularity of gait initiation is it poses important challenges to balance. In walking, each limb contributes approximately in the same manner to forward progression. Conversely, in gait initiation, the roles of each limb are quite distinct and under different command [55]. Indeed, in quiet standing, prior to the APA's and gait initiation, each limb is initially loaded to about 50% of the BW. When initiating gait, all BW must be placed onto one limb, (i.e. trailing limb) freeing the other limb (i.e. leading limb) and allowing it to swing forward for the first step to be made. In doing so, important challenges to balance are posed. In fact, placing the entirety of the BW on one limb necessitates sufficient strength and control in the trailing limb. As described above, to produce a forward movement of the body CoM, in keeping with the model of the inverted pendulum outlined above, the CoP_{net} must move posteriorly in order to destabilize and push the CoM forward via the APA's [42].

Gait initiation has been well described in healthy adults and numerous studies have investigated the APA's of the CoP_{net} and CoM interaction [41,52,55,56]. The APA's in gait initiation have been defined by four distinct phases [106]. These APA's are related to the displacement of the CoP_{net} in both the A/P and M/L directions and take place in the following sequence: 1) APA1 is a displacement of the CoP_{net} posteriorly and laterally toward the leading limb; 2) APA2 is a medial and slightly anterior displacement of the CoP_{net} towards the trailing limb, as BW is loaded to the trailing limb. The end of the second phase occurs when the CoP_{net} is approximately centered between both limbs and there is heel-off of the leading limb; 3) APA3 is a posterior and lateral displacement towards and beneath the trailing limb, which takes place as there is leading limb toe-off; and finally, 4) APA4 is a rapid forward displacement of the CoP_{net} with the trailing limb toe-off, the CoP_{net} travelling from heel to toe-off. Indeed, the APA's are referred to as anticipatory (i.e. prior to gait initiation) but they continue through to toe-off of the trailing limb [56]. A sketch of the APA's CoP_{net} trajectory is outlined below in Figure 8.

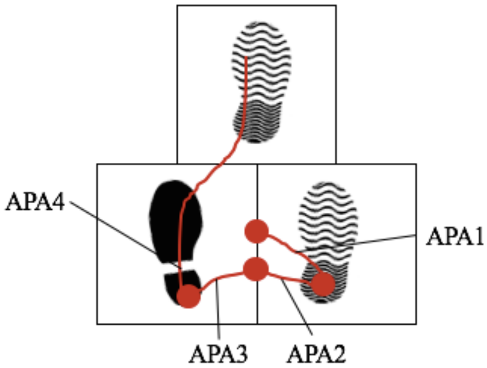


Figure 8- CoP_{net} displacement in the four phases of APA's involved in gait initiation. Left foot (solid black) is the trailing limb while the right foot (gray outline) is the leading limb, initiating the first step. Four distinct APA phases (APA1-APA4) and associated CoP_{net} displacement (red line) are shown. APA1 is a displacement of the CoP_{net} posteriorly and laterally toward the leading limb; APA2 is a medial and anterior displacement of the CoP_{net} towards the trailing limb; APA3 is a posterior and lateral displacement towards and beneath the trailing limb and finally, APA4 is a rapid forward displacement of the CoP_{net} on the trailing limb.

During the sequential APA phases described above, the lateral displacement of the CoP_{net} allows the leading limb to be unloaded and total BW is transferred to the trailing limb (i.e. displayed as left on Figure 8). Simultaneously, the backward CoP_{net} displacement *pushes* the CoM forward as the leading limb leaves the ground. The efficacy of this CoM push is related to the body configuration and the CoP_{net} position. When standing upright, approximately 2/3 of total BW is carried by the head, arm and trunk segments and the associated CoM is also located at approximately 2/3 of the height of the individual. This geometry makes the inverted pendulum inherently unstable [42]. During gait initiation, the subjects take advantage of this biomechanics, moving the COP_{net} backward which pushes the CoM in a forward progression [42].

After the initial APA phase, gait initiation is also produced by the application of forces on the ground generated by movements of the lower limb segments. Thus, the spatiotemporal, kinematic and kinetic parameters during gait initiation will be presented in the following paragraphs.

During quiet standing, the CoM velocity is negligible, and through gait initiation, CoM accelerates and reaches SSWV. This SSWV is specific to each individual in order to minimize the energy expenditure [15,27], and reach a SSWV of approximately 1.3 to 1.5 m/s with a very low intra-individual variability in healthy adults [57,58,59,60]. That is, greater physiological and energetic demands are present at slower or faster walking velocity than that of SSWV [61,62].

Spatiotemporal parameters, such as cadence (number of step per minutes), step and stride length have seldom been studied with regards to gait initiation [56,63,64]. Most spatiotemporal parameters arise from SSWV as is seen in equation 3.

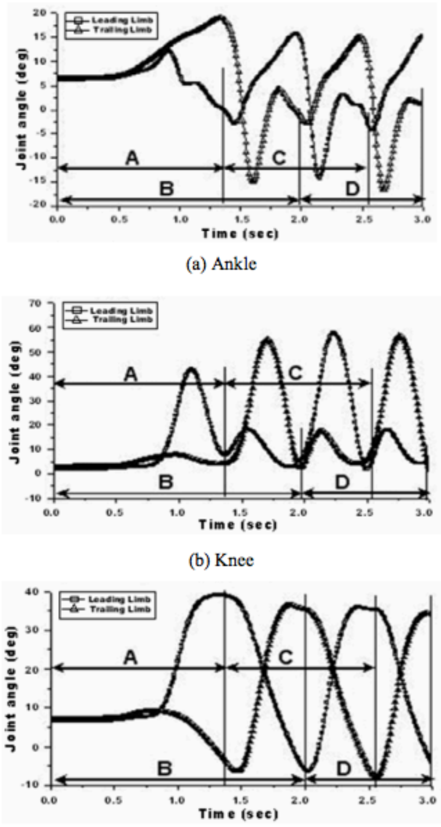
$$\text{walking velocity (m/s)} = \frac{\text{cadence (step/min)} \times \text{stride length (m)}}{120} \qquad \text{(Equation 3)}$$

SSWV is thus an important encompassing parameter. Therefore, the spatiotemporal parameters of the first step of gait initiation are modulated according to the speed at which gait initiation is performed [65]. Most studies investigating gait initiation have focused solely from the APA through to toe-off of the trailing limb, without interest for the underlying biomechanics of the first step of gait initiation, thus omitting spatiotemporal parameters other than SSWV [42,53]. Hence, fewer studies have evaluated the biomechanics of the whole process of gait initiation, providing data with regards to the spatiotemporal parameters of the first step(s) in gait initiation [56,63,64].

Park et al. found that the step and stride length, along with walking velocity, gradually increased from gait initiation until SSWV was reached, the main difference occurring between quiet standing position and the first step. For example, they reported that the step length was 0.40 to 0.60 m in the first step made with the leading limb as opposed to a step length of 0.70 m reached during SSWV. After the first step, all spatiotemporal parameters were said to approach and resemble that of SSWV [110].

Kinematics are those biomechanical parameters which describe movement, without concern for the forces which produce it [43]. In gait analysis, kinematics typically describes the position, velocity and the acceleration of segments and joints when they are moving in both linear and angular 3D space. Park and colleagues documented joint angle kinematics in 20 healthy male adults [110]. Figure 9 below outlines the average sagittal joint angles throughout 15 gait initiation trials, from the APA through to SSWV. The results of the ankle joint kinematics of the leading limb showed a plantarflexion of 12.7° (compared to about 20° in healthy adults during SSWV) and the trailing limb showed 19.8° at heel-off [66]. For the leading limb, from quiet standing, through APA, to step 1 (Interval A in Figure 9), the maximum knee flexion was reduced and the maximum hip flexion was increased when compared to SSWV.

As with the spatiotemporal parameters, the joint kinematics progressively resembles that of SSWV with each additional step from gait initiation. Moreover, the first step is involved with the most important changes in producing joint kinematics which resemble that of SSWV, each additional step thereafter producing minor adjustments until SSWV.



(a) Ankle

(b) Knee

(c) Hip

Figure 9- Sagittal plane joint angles of the (a) ankle, (b) knee and (c) hip during gait initiation. Average profiles for 15 gait initiation trials in 20 healthy male adults. Interval A represents the gait initiation to heelstrike in the leading limb and interval C is the second step in the same limb. Interval B represents gait initiation to trailing limb heelstrike and interval D represents the second step of the trailing limb. (Adapted from Park et al. 2009 [110])

Kinetics are those biomechanical parameters which describe the forces that produce movement [43]. Of many various kinds, these parameters are most often divided between CoP and forces. The GRF are those forces equal and opposite to the forces applied to the ground[43]. GRF are inputs for CoP calculation and are typically measured via force plates in all three directions: A/P, M/L and vertical. These provide important insights into how BW is loaded and unloaded (via vertical forces) and how the CoM is accelerated in the horizontal plane via A/P and M/L forces. GRF can also be quantified with regards to the peak force applied, loading rate (speed of force applied) and their associated temporalities. In order to compare between individuals, GRF (measured in N) are typically normalized to mass of the subject (N/kg).

The vertical GRF of each limb is reflective of the fraction of BW applied to the limb. During gait initiation, and as shown previously, the leading limb is unloading the vertical force while total BW is transferred to the trailing limb. Then, the vertical force on the trailing limb increases to full BW while the leading limb moves forward to initiate the first step. Figure 3 below shows the vertical GRF observed in the leading and trailing limb in a healthy female subject [53]. The values of vertical GRF observed during the APA phase (interval S to R, Figure 10) demonstrate the shared BW between both feet. Prior to the leading limb toe-off, during the APA1, as the CoP_{net} is displaced to beneath the leading limb, the vertical GRF reach values of approximately 80% BW (phase M, Figure 10). Then, at toe-off of the leading limb, the BW becomes null and all BW is rapidly loaded to the trailing limb. The CoP_{net} has then travelled to a point posteriorly beneath the trailing limb (end of APA 3), and the trailing limb reaches a vertical GRF of about 110% BW. When the leading limb contacts into step 1, double support is therefore present and there is a partial transfer of BW from the trailing limb to the leading limb. Finally, with the action of the ankle joint plantarflexion force generation, and as the CoP_{net} travels forward in APA4 to toe-off, there is another peak vertical GRF reaching about 120% BW (Max3, Figure 10) during push off before the trailing limb leaves the ground [67]. The vertical GRF of the first step in gait initiation have less been studied.

Two important mechanisms produce forward movement during gait: the interaction of the CoM and CoP_{net} (inverse pendulum) and the power generation produced at the ankle [41,67,69]. The ankle plantarflexors are responsible for force generation and forward propulsion during gait initiation [6,28,67]. These important force generators can be seen here as contributing to the second peak of vertical GRF (Max3, Figure 10) before trailing limb toe-off as well as to that of the subsequent steps (not outlined in Figure 10).

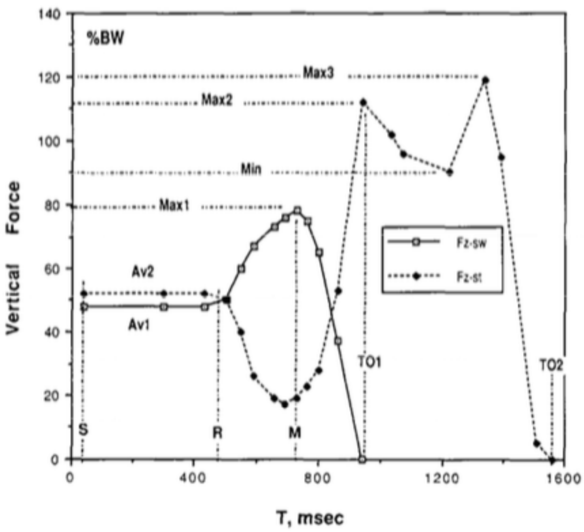


Figure 10- The vertical GRF profile during gait initiation. Average profile in one healthy female during gait initiation. Quiet standing, prior to APA's, is outlined by the interval S to R. Peak leading limb vertical force is outlined at Max1 and peak push-off of the trailing limb is outlined as Max3 (adapted from [53]).

The A/P forces are those which describe the braking (anterior) and propulsive (posterior) components. In gait initiation, there is a change from a static CoM to a forward CoM progression. This can be produced by applying propulsive forces beneath the leading and trailing limbs. The M/L forces are also applied during gait initiation but they will be omitted from the current chapter as they do not contribute to forward propulsion and the CoP_{net} displacement in the M/L direction has been discussed with regards to APA's. During the first step, as walking velocity is increasing, propulsive force must be greater than the braking force [68]. As shown in Figure 11, the A/P forces are representative of the APA's CoP_{net} displacement. Quiet standing (interval S to R,

Figure 11) shows negligible A/P GRF. Then, as the leading limb pushes off the ground to toe-off, there is an increase in propulsive force reaching about 10% BW. During this, minimal A/P GRF are generated by the trailing limb. Then, prior to toe-off, the trailing limb quickly develops important propulsive forces to approximately 25 to 30% BW.

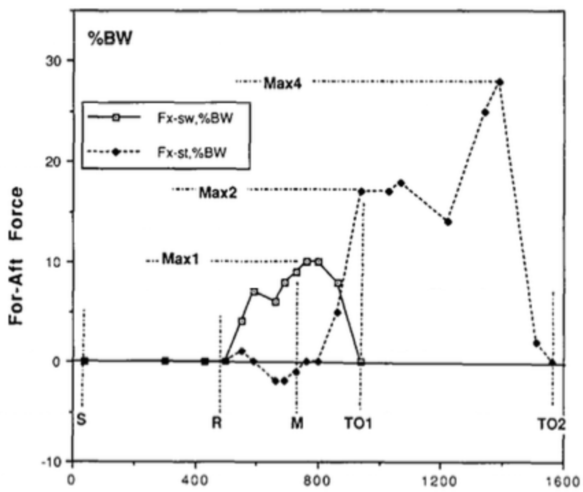


Figure 11- The A/P GRF profile during gait initiation. Average A/P GRF in one healthy female during gait initiation. The forces are positive when pushing the body forward. The interval S to R denotes quiet standing prior to APA. The maximum propulsive force produced by the leading limb at push-off is outlined at Max1. Max4 outlines the push-off force of the trailing limb (adapted from Nissan & Whittle (1990) [53]).

Although three steps are typically needed to reach SSWV, minimal increase of walking velocity are achieved by the second and third step [53,69,70]. That is, the adjustments made to spatiotemporal, kinematic and kinetic parameters are minor and thus the first step of gait initiation is the most important in producing SSWV [42,53,69,70,110].

3.1 Gait Initiation in Transtibial Amputees

With regards to the gait initiation pattern, fewer studies have been conducted in the TTA. In this section, gait initiation in the TTA will be presented with regards to specific studies in order to explore the different methodologies used to measure the spatiotemporal, the kinematic and the kinetic parameters.

A first study conducted by Nissan (1991) investigated gait initiation in ten unilateral TTA (causes of amputation were not provided). Kinetic data was provided from beneath each foot during quiet standing and until final push-off of the trailing limb. Participants initiated gait with their preferred limb when a visual cue was given and kinetic and kinematic data were recorded. Interestingly, with regards to preferred leading limb, half the TTA (n=5) chose the prosthetic limb, while the other half (n=5) chose to initiate gait with the intact limb. The average of three gait initiation trials was tabulated. With regards to kinematics, the results of range of motion at the hip, knee and ankle were reduced in both limbs, with the exception of an increased hyperextension of the hip trailing limb when compared to controls. The kinetic parameters examined showed significantly reduced peaks for propulsive A/P and vertical forces (i.e. all recorded peak forces). As well, a smaller active plantarflexion push-off by the prosthetic trailing limb was observed and this was expected due to the passive prostheses worn by participants. The author concluded by stating that the TTA showed a tendency for a slower and more careful gait initiation pattern. Indeed, the reduced plantarflexion force produced by even the intact trailing limb and along with that of the prosthetic limb was evidence of this. Unfortunately, no kinetic data was recorded on the first step [71].

A second study with regards to gait initiation in TTA was carried out to characterize the CoP_{net} displacement and GRF [72]. Seven TTA subjects, aged 50 to 82 years old, participated in this study and no control group was included. Participants were provided with the same prosthesis and initiated gait upon visual cue. Therefore, this study, to some extent, observed the change in gait initiation profile brought about by a change in prosthesis. Subjects were asked to complete three gait initiation trials with each limb leading. Kinetic data was recorded beneath each foot during quiet standing through to trailing limb toe off (i.e. end of APA3 phase). Though it was not named as such, the CoP_{net} displacement results followed the APA sequence (from phase 1 to 3) as outlined in the healthy adults gait initiation section above. Indeed, an overall posterior CoP_{net} shift was observed prior to gait initiation and the authors concluded that this displacement of the CoP_{net} warranted further investigation.

With regards to GRF, during quiet standing as in gait initiation, the TTA consistently loaded the intact limb more than the prosthetic limb. This was even true when the TTA were in single stance support as time spent on the prosthetic limb was reduced and time spent on the intact limb was increased. This is well seen in Figure 12 below, which outlines the average vertical GRF between the prosthetic and intact limbs for all seven subjects across all 6 trials. All peak forces are significantly reduced in the prosthetic limb when compared to the intact limb. The preferred limb for gait initiation and the cause of amputation were not provided and no kinematics were reported.

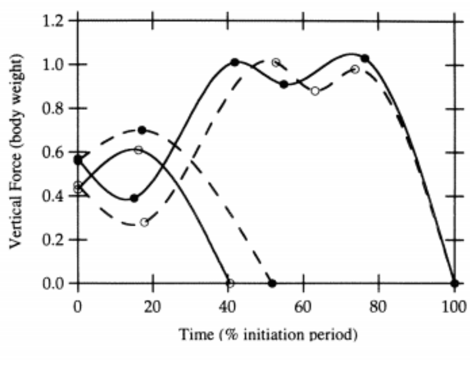


Figure 12- Vertical GRF in the prosthetic and intact limb during gait initiation. The average vertical GRF profiles in seven TTA across 6 trials in prosthetic (hollow

dots) and intact limb (full dots) with the leading prosthetic limb (solid line) and leading intact limb (dashed line) conditions (Adapted from Rossi et al. 1995 [72]).

Tokuno and colleagues, in 2003, compared gait initiation in 11 TTA (8 traumatic, 2 dysvascular and 1 cancer) and 11 age-matched control subjects. Following the results of Rossi and colleagues, Tokuno et al. (2003) were also interested in understanding the CoP_{net} displacement prior to and during gait initiation and they hypothesized that the intact limb would exhibit temporal, kinematic and kinetic compensations compared to the prosthetic limb.

Participants initiated gait from a quiet standing position upon cue from a light, which indicated which limb to initiate gait with. Subjects stepped forward onto a third force plate placed at a percentage of their preferred step length (+0%, +25% or +50%). To some extent, this study therefore also investigated the effect of step length in gait initiation as well as motor control as step length and leading limb were randomized throughout trials. Fourteen trials were completed at each step length and kinetic data was collected from three force plates.

It was observed that the TTA preferred to initiate gait with their prosthetic limb in 8 of the 11 TTA. Gait initiation required more time compared to controls, regardless of leading limb. The results obtained with regards to CoP_{net} displacement showed large differences between intact and prosthetic limb leading conditions. The posterior CoP_{net} displacement in the intact limb resembled that of controls, but the posterior CoP_{net} displacement was greatly reduced beneath the prosthetic limb.

With regards to GRF, solely the A/P forces were described for force plates 1 and 2. When leading, the prosthetic limb produced significantly reduced braking and propulsive peak forces when compared to the intact limb. When trailing, no difference was observed between the prosthetic and intact limbs while both limbs of amputees showed reduced braking and propulsive peak forces compared to controls. For force plate 3 (step 1), both braking and propulsion forces were significantly reduced in the prosthetic limb when compared to the intact and control limbs. The first step vertical forces were also reported and significant differences existed between the prosthetic and intact limbs. The two vertical GRF peaks at weight acceptance and push-off were both reduced in the prosthetic limb when compared to the intact limb. No significant differences were observed between controls and both TTA limbs as forces were non-significantly increased in the intact limb and non-significantly reduced in the prosthetic limb when compared to controls. However, the midstance minimum observed in vertical GRF was further reduced in the prosthetic limb when compared to both the intact and control limbs.

Impulse is an additional kinetic parameter by which to quantify components of gait initiation. Indeed, it is the integral of the force curve with respect to time (t) and it is measured in N·s/kg:

$$Impulse = \int F(t)dt \qquad \text{(Equation 4)}$$

Tokuno and colleagues also calculated the A/P impulse (braking and propulsive) during quiet standing and APA through to toe-off. A significant reduction in the prosthetic limb A/P impulse (both braking and propulsive) when compared to intact limb was seen, regardless of leading limb condition. Authors described a ‘movement time’ strategy employed by the TTA, favoring stability over propulsion, like the careful gait initiation strategy described by Nissan and colleagues. This was the first study to investigate the GRF involved beneath the first step but no impulses were calculated from this first step [55].

Michel and Chong (2004) conducted a study with regards to gait initiation in 5 TTA and 6 transfemoral amputees when compared to two able-bodied subjects. They examined the mechanisms used to control the propulsive forces in order to regulate the forward velocity during gait initiation. Participants were asked, upon auditory cue, to initiate gait in four conditions (normal and fast speeds, and with prosthetic or intact limbs) for a total of 10 trials per condition. All participants initiated gait from bilateral stance in quiet standing on a single force plate. Their results indicate that the TTA reached the same velocity, regardless of leading limb (i.e. prosthetic or intact) in the normal speed condition. With regards to temporality, the TTA increased the time from loading to the toe-off in the leading limb, regardless of leading limb (prosthetic or intact). This confirmed the results of Tokuno and colleagues, stating that the increased time in gait initiation was used to create the propulsive impulse [55]. Again, Michel and colleagues put forth the idea of reduced propulsion to augment stability in the TTA. Limits of the study include that only a single force plate was used to record data in the TTA group. As well, many assumptions with regards to unilateral amputees were based on the dual platform trials carried out in two of the transfemoral amputees, without actual testing in the TTA [73].

Miff and colleagues (2005) investigated the temporal characteristics of the CoM acceleration during gait initiation and gait termination in ten TTA (trauma n=6; cancer n=2; infection n=2) when compared to ten able-bodied individuals. Participants were asked to complete three gait initiation and three gait termination trials, with each limb, at slow, normal and fast speeds. Six kinematic parameters were observed and data was collected using a motion analysis system solely. Therefore, APA times were calculated based on the events of the CoM (i.e. start of acceleration of CoM to toe-off of the leading limb representing APA1 and APA2 phases). Results suggest that approximately two steps are needed to reach SSWV in both amputee and control groups, and that approximately all SSWV is reached with step 1 (i.e. 70%). The results also demonstrated no difference in time to reach the SSWV when gait was initiated with either the intact or prosthetic limbs. Unfortunately, controls were not age-matched to the TTA group and so effects due to age interaction are possible (i.e. SSWV) [74].

Vrieling and colleagues conducted two studies with regards to gait initiation in unilateral TTA. In both protocols, gait was self-initiated rather than following a visual or auditory cue [76,77].

In a first study, Vrieling and colleagues investigated the kinetics related to gait initiation in TTA, as results of prior studies, with regards to the CoP_{net} displacement during APA, varied (i.e. Rossi et al. 1995, Michel & Chong 2004, Tokuno et al. 2003). Twelve TTA (trauma n=6, vascular n=2, cancer n=4), 7 transfemoral amputees and 10 control subjects were asked to self-initiate gait with both their intact and prosthetic or right and left limbs. All participants completed 8 trials on a walkway where leading limb preference and single limb stance duration time were assessed via video cameras. As well, 4 more trials were collected initiating gait from a single force plate in order to obtain the GRF, the CoP displacement and the gait initiation velocity. In these force plate trials, participants had to alternate right and left or intact and prosthetic as leading limb. Results indicate that most often (8 of 12 TTA) the prosthetic limb was the preferred leading limb and no difference was observed in gait velocity when intact or prosthetic side was the leading limbs. Single limb stance time in the trailing prosthetic limb was reduced when compared to the trailing intact limb.

With regards to the GRF pattern collected from the force plate beneath both feet, the vertical weight acceptance peak in the leading prosthetic limb was reduced when compared to controls and intact leading limb conditions. For the final vertical peak force of the trailing limb, a decrease in force was also observed in the prosthetic trailing limb condition when compared to controls and intact trailing limb conditions. The A/P propulsive force created by the intact and prosthetic leading limb was decreased when compared to controls. The A/P propulsive force beneath the trailing limb was also decreased in the TTA in both limbs when compared to controls. Finally, the A/P propulsive force beneath the prosthetic trailing limb was decreased compared to intact trailing limb condition.

In the leading intact limb condition, during the APA, there was an increased M/L CoP_{net} displacement from the leading limb towards the trailing prosthetic limb in the TTA when compared to control and intact trailing limb conditions. For the A/P CoP_{net} displacement beneath the leading limb (APA1), there was a reduced posterior CoP_{net} displacement observed in both limbs of TTA when compared to controls. As well, a reduced posterior CoP_{net} displacement was observed beneath the prosthetic limb when compared to the intact limb. For the A/P CoP_{net} displacement beneath the trailing limb (APA3), the CoP_{net} displacement was anterior beneath the trailing prosthetic limb when compared to a posterior CoP_{net} displacement observed in controls and the intact trailing limb condition.

With regards to the strategies utilized by the TTA, Vrieling et al. found that propulsive forces were reduced in the prosthetic (and preferred) leading limb condition and there was no increase in the intact trailing limb propulsive forces. Instead, a prolonged duration of single limb stance existed in the intact trialing limb. Thus, a larger propulsive impulse was created in the trailing intact limb, a compensatory mechanism. They also observed that the propulsive forces of the trailing limb were reduced in the TTA groups and for two following reasons: 1) restricted initial posterior displacement of the CoP_{net} (i.e. reduced posterior and even anterior CoP_{net} displacement beneath the trailing limb APA3 phase), and 2) the absence of the ankle plantarflexors in the prosthetic limb. Again, authors conclude of a want for stability exhibited in the TTA. The limits of this study pertain to the equipment and experimental protocol. Only one force plate was used to record kinetics beneath the feet during gait initiation providing limited data and approximations of the various APA phases. Finally, the authors note that gait initiation kinetics after initial toe-off and subsequent steps are warranted [75].

The second study by Vrieling and colleagues, published a year later, was carried out in 7 TTA, 4 transfemoral and 3 knee disarticulation amputee population during the rehabilitation process. Reasons for amputation were vascular disease (n=12), tumor (n=1) and infection (n=1). Unfortunately, due to several complications, 7 participants were not able to participate in 1 or more of the 4 assessments of the study. One force plate was embedded along a walkway, recording kinetic data beneath both feet during the bilateral stance phase prior to gait initiation. Kinematics was recorded via the use of video cameras and electrogoniometers fixed at the hip and knee joints. Although this study investigated the effect of a rehabilitation program in various levels of amputation, as well as the use of various prosthetic devices, results were pooled together, though authors noted greater ability in the TTA participants. The results confirmed a reduction of posterior CoP_{net} displacement beneath the trailing prosthetic limb when compared to able-bodied individuals, as previously reported [72]. Limits of this study include the large number of drop-outs, the presence of waking aids for some trials depending on the participant's stage of rehabilitation as well as the lack of distinct groups for levels of amputation [76].

As stated by Vrieling and colleagues, gait initiation requires two skills which are limited in the TTA: propulsion and balance control [75]. As well, the strategy to favour stability over propulsion in gait initiation is well supported by the current TTA gait initiation literature. As first step following gait initiation provides the greatest contribution to SSWV, understanding of the kinetics of the first step are needed, as much as those related to APA.

Thus, to resume, the gait initiation studies carried out thus far in the TTA have shown a reduced posterior or anterior A/P CoP_{net} displacement during the APA phase. The prosthetic limb has been shown to be the preferred leading limb for gait initiation as this is a more stable position. The range of motion of the hip, knee and ankle are reduced in both the prosthetic and trailing limb, with the exception of hip hyperextension seen in the trailing limb, regardless if intact or amputated. Finally, the kinetic strategy utilized by the TTA is to increase the time of force application in order to compensate for the reduced force produced by the prosthetic limb, though all propulsion remains decreased in both the intact and prosthetic limbs when compared to controls. All these strategies employed by the TTA in gait initiation support the idea of a slower and more careful pattern, favouring stability over propulsion.

With regards to cause of amputation, as mentioned, a number of studies have recently undertaken gait analysis by considering the traumatic TTA and DTTA as two distinct populations [27,29,30,77,78]. Indeed, the various studies have investigated both SSWV and quiet standing and significant differences have been observed between both groups. However, in gait initiation studies thus far conducted, no distinction has been made with regards to the cause of amputation. Knowing that the DTTA has important additional confounding conditions (i.e. sensory loss) which have been shown to impact quiet standing stability and SSWV, it is plausible that strategies employed to initiate gait would also be unique in the DTTA. Yet no studies have explored gait initiation in the specific DTTA population. Hence, studies of gait initiation in the specific DTTA population are warranted. Thus, the second and third purpose of the current thesis are to characterize the APA and first step kinetics strategy employed by DTTA when compared to their age-matched controls.

Objectives & Hypotheses

Objective I: The first objective of this thesis is to determine the most relevant biomechanical parameters used for gait analysis in a healthy adult population. The term relevant is defined as those biomechanical parameters being able to identify gait abnormalities in the healthy adult population and applicable to the clinical and rehabilitation setting.

Hypothesis I: Through a systematic review of the literature, it is hypothesized that those biomechanical parameters most relevant to gait analysis in the healthy adult population would be walking velocity. As well, it was hypothesized that Level of Evidence, Impact Factor and Relevance Score would be positively correlated with the most relevant parameters.

Objective II: The second objective of this thesis is to compare the anticipatory postural adjustments during gait initiation used by dysvascular transtibial amputees with those of age-matched controls.

Hypothesis II: It is hypothesized that the anticipatory postural adjustments used by the dysvascular transtibial amputees will exhibit a reduced total posterior CoP_{net} displacement beneath the prosthetic trailing limb in the dysvascular transtibial amputees when compared to the healthy controls. As well, it is hypothesized that increased total APA time will be observed in the dysvascular transtibial amputee when compared to age-matched controls.

Objective III: The third objective of this thesis is to compare the underlying biomechanical differences in the first step kinetics of dysvascular transtibial unilateral amputees with those of healthy age-matched controls.

Hypothesis III: It is hypothesized that the dysvascular transtibial amputees will show reduced propulsive impulse during gait initiation in both the prosthetic and intact limbs, when compared to the healthy controls. As well, it is hypothesized that, in the dysvascular transtibial amputees, the intact limb will show significantly greater propulsion impulse when compared to the prosthetic limb, as a compensation mechanism.

Chapter 2: **Methods**

Objective 1

For the first purpose of this thesis, the systematic review excluded studies with participants living with pathologies, disabilities, health concerns and/or neurological deficits were excluded. To be selected, articles had to evaluate adults aged 18 to 65 years old with no walking aids.

An online search in three databases (i.e. PubMed, EMBASE and Web of Science) was carried out for the systematic literature review. These three databases were selected for search because of their broad inclusion of multidisciplinary topics within the Biomedical and Health Sciences domain. Each database was searched for all years included in the respective databases with the last search completed in May 2016. In databases where applicable, certain additional parameters were used to narrow the search. In PubMed, filters including human studies of adults aged 18 to 65 years old, published in French and English and with regards to the nature of the study (i.e. original articles, review articles, case study) were applied. In the EMBASE and Web of Science databases, filters were applied to include human studies, French and English language publications and specific nature of study (i.e. original articles, review articles, case study).

A census of all biomechanical parameters measured was undertaken by two evaluators by carefully reading and analyzing the chosen articles for the systematic review. First, all methodological aspects of the selected articles were tabulated and briefly summarized. Second, the biomechanical parameters measured in all articles were tallied. Third, because of the many various instruments, techniques, planes of measurement, etc. used to quantify parameters in the studies selected, the parameters measured were summarized under broader parameter names (i.e.: sagittal, frontal and transverse plane knee power were combined under the broader name of knee power). Lastly, after a summation of parameters, the number of different articles measuring a type of biomechanical parameter was counted; this was also done for single parameters.

In an attempt to evaluate relevance of biomechanical parameters, both the frequency of measure and the number of different articles which measure the parameter were combined to produce a score using the summarized parameters. We evaluated quality of the selected articles by attributing a Level of Evidence score for each selected article. Our Level of Evidence score was a modified version of that used by Sagawa and colleagues [105], since they were interested in the gait analysis in a population with a lower limb amputation and the current systematic review addresses healthy adult gait analysis. The 14 criteria were subdivided between three main article elements: 1) selection of participants, 2) intervention and assessment, and 3) statistical validity. The maximum possible score was therefore 14, with each article receiving a score of 1 (if they met the requirements) or 0 (if they did not meet the requirements) for each criterion (score of 1 for a non-applicable criterion). Two independent evaluators assessed the score of all articles. For any disparities between scores, both evaluators determined the best suited scoring through discussion. If a consensus could not be reached by the two evaluators, a third evaluator intervened in order to break tie between both scores suggested.

The outcome parameters of this systematic review carried out in healthy adults, as outlined by the above procedures, included frequency of measurement of biomechanical parameters, number of articles measuring a given parameters, Level of Evidence score, Relevance score as well as the Impact Factor of the journal the year the article was published.

Statistical analysis was carried out using Spearman correlations in order to determine if higher Level of Evidence articles are published in higher Impact Factor journals, as well as, to determine the relationship between the mean Level of Evidence attributed to all articles measuring a given parameter and its frequency of measurement. All statistical analyses were carried out using SPSS 22 (IBM Corp., NY). Level of significance was set at $p < 0.05$.

Objectives II & III

For the final two purposes of the current thesis, with regards to APA and first step kinetics of gait initiation, a total of 10 subjects with a unilateral DTTA were recruited via the Institut de réadaptation Gingras-Lindsay de Montréal. A group of 10 control subjects were recruited via acquaintances of the researchers at l'Université de Montréal. The control subjects were healthy adults, age-matched to the DTTA subjects.

Conditions and/ or diseases which could have an impact on the standing and locomotor pattern (i.e. other than that having caused amputation, for example Type II Diabetes) were reason for subject exclusion. All DTTA subjects wore their own prosthetic device for testing and all prostheses were equipped with a passive foot. All subjects provided informed consent prior to testing. This study was approved by the Comité d'éthique de la recherche en santé de l'Université de Montréal and the Comité d'éthique en recherche des établissements du Centre de recherche interdisciplinaire de réadaptation du Montréal métropolitain.

A walkway, with three embedded AccuGait force plates (Advanced Medical Technology Inc., MA), was set up surrounded by 8 Flex13 motion capture cameras from the OptiTrack motion analysis system (NaturalPoint Inc., OR). A total of 39 reflective markers were placed on the subject at anatomical landmarks based on the Plug-in Gait model (Vicon Motion Systems Ltd., UK). Subjects were asked to initiate gait and walk looking straight ahead during each trial to avoid targeting of the force plates. Practice was allowed in order to ensure targeting did not take place. The force plates measured the GRF and moments in all three planes of movement (vertical, anteroposterior (A/P) and mediolateral (M/L). Both kinetic and kinematic systems were synchronized and sampled at 100Hz. All data analysis was carried out using a MATLAB program (The MathWorks Inc., MA) created for the purpose of the research projects.

Subjects were asked to change into their athletic attire and reflective markers were then fastened to skin, prosthesis and/or tight-fitted clothing. Subjects were first asked to self-initiate gait (i.e. no start cue was given) with their right limb, from quiet standing with each foot on force plates 1 and 2, naturally stepping onto the third force plate with their first step and continuing to the end of the walkway. Subjects were then asked to initiate gait with their left limb in the same manner. Five trials with each limb leading were collected. Subjects were informed that they could rest at any time and all improper trials were deleted and collected once again.

For the purpose of the APA in the DTTA when compared to healthy controls, data collected from the three force platforms was exported and the CoP_{net} was calculated across all three force platforms. Parameters of CoP_{net} A/P displacement in cm (A1, A2, A3), CoP_{net} M/L displacement in cm (M1, M2, M3) and duration of phase in seconds (T1, T2, T3) for each APA phase were calculated. As well, the total APA CoP_{net} A/P displacement in cm, CoP_{net} M/L displacement in cm and duration in seconds (A_{total}, M_{total}, T_{total}) were calculated from quiet standing to the end of APA3 phase and tabulated. The APA4 phase describes the end of the APA, as the trailing limb leaves the ground, and the CoP_{net} pattern of APA4 is omitted from the current analysis. The main outcome parameters were established as A_{total} and T_{total}. Thus, with an error alpha of 0.05, a power of 80% and minimal difference to detect of 0.14s for T_{total} and 1.54cm for A_{total}, a sample size of 25 participants were needed per group.

For the final purpose of the current thesis, from the kinetic and kinematic data collected, the heel contact and toe-off events were identified and gait velocity was calculated. The maximum braking, propulsive and vertical forces as well as associated impulses and loading rate of the first step were computed. Impulses were calculated as the time-force integral and loading rate as the slope of the force (i.e. force divided by time) leading to the maximum vertical force. Force and impulse parameters were divided by body weight and time normalized to 100% of stance phase. The main outcome parameters were established as SSWV and propulsive impulse. Thus, with an error alpha of 0.05, a power of 80% and minimal difference to detect of 0.16m/s for SSWV and 0.48N•s/kg for propulsive impulse, a sample size of 25 participants was needed per group.

For the second and third objectives of the thesis, a non-parametric analysis of variance was carried out via a Friedman related samples test with a Bonferroni correction. A pairwise comparison was done in order to control for age. Further, Wilcoxon related sample tests were carried out between conditions for each parameter (i.e. control, intact limb, prosthetic limb). Statistical design compared both conditions (i.e. gait initiation with the right or left leading limb) in control subjects. For instances when the right and left leading limb conditions were not statistically different in controls, the mean result of both limbs was taken as one control limb result. In DTTA subjects, gait initiation with the prosthetic versus intact limb was compared. Finally, the results obtained in the control group were compared to those obtained in the DTTA group (i.e. control, intact limb, prosthetic limb).

Effect size was also calculated in order to observe clinical significance of all measured parameters between the control, intact and prosthetic limb conditions. Cohen's d was used to the calculate effect size ($d \geq 0.5$ = moderate clinical significance and $d \geq 0.8$ = strong clinical significance for those parameters significantly different statistically).

All statistical analyses were carried out using SPSS 24 (IBM Corp., NY) at a level of significance of $p < 0.05$.

Introduction of Articles

The three articles hereafter presented in this thesis propose to document the established purposes of the current thesis. It was thought most appropriate to present this thesis in article format in order to accelerate the dissemination of results.

The first article, a systematic review of the literature pertaining to healthy adults gait analysis, will consider the first purpose of the current thesis in determining the most relevant biomechanical parameters used for gait analysis in a healthy adult population. This systematic review was published in *Physical Therapy and Rehabilitation* (2017) 4: 6.

The second article will aim to answer the second objective of this thesis, to compare the anticipatory postural adjustments during gait initiation used by dysvascular transtibial amputees with age-matched controls. This article is to be submitted to Gait & Posture for publication.

The third and final article, will compare the first step kinetics of gait initiation in the DTTA with healthy controls as a means to answer the third purpose of this thesis. This article is to be submitted to Gait & Posture for publication.

Chapter 3: Article I

Biomechanical parameters for gait analysis: a systematic review of healthy human gait.

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Abstract

Background: Modern gait analysis offers a broad variety of biomechanical parameters through which to quantify gait. However, no consensus has yet been established with regards to which biomechanical parameters are most relevant to evaluate during gait analysis in the healthy population. Purpose: The purpose of the current systematic review was to determine the most relevant biomechanical parameters for gait analysis in the healthy adult population. Methods: PubMed, EMBASE and Web of Science databases were searched. Two independent reviewers participated in the article selection and attributed a Level of Evidence score to each article to account for quality based on participant selection, intervention and analysis. A score combining both frequency and number of articles was calculated Correlations were carried out between the Level of Evidence score, Journal Impact Factor and the frequency of biomechanical parameters. Results: Spatio-temporal parameters were found to be the most often measured biomechanical parameters and reported by the greatest number of articles, walking velocity, cadence and step/stride length appearing to be the most relevant biomechanical parameters for gait analysis in the healthy adult population. No correlation was found between Level of Evidence score and Journal Impact Factor, nor between the frequency of parameters and Level of Evidence score. Conclusion: This systematic review provides recommendations for variables to assess in future gait evaluations in healthy adults.

Keywords: Gait, Biomechanics, Gait Analysis, Healthy, Adult.

1. Introduction

Walking is the most common form of locomotion and it is part of almost all activities of daily living [1,2]; therefore, the ability to walk is an indicator of overall health as it dictates autonomy [3]. Although walking is usually learned at a young age, the mechanics of walking are not as simple as they may appear [1].

From the first studies of human walking elaborated through a series of photographic images, by early Biomechanics enthusiasts Edweard Muybridge and Étienne-Jules Marey, gait analysis as it is known today has evolved significantly [4]. The walking pattern of individuals has become an area of broad interest and the focus of much research as seen by the numerous journals and articles published. The importance of gait analysis lies in its application; through years of research and experimentation, gait analysis has become widely used as a means to diagnose pathology, set a prognosis and establish and evaluate a treatment plan [5,6]. Today, a variety of different parameters of various types exist and are readily used to examine and explain human gait [7,8,9,10].

In clinical settings, gait analysis is often carried out solely through clinician observation[11]. Although clinicians have developed good expertise through many years of practice and training, these observations remain subjective [12]. Principal reason for main, and perhaps sole use of clinician observation as means of gait analysis, is ease of measurement [8,13,14].

In the research setting, numerous parameters have been used to quantitatively describe gait. Parameters of various types such as spatio-temporal parameters, ground reaction forces, joint kinematics and the energy expense are a few [1,15,16].

In accordance with evidence-based-medicine, the biomechanical parameters chosen are as important as rigorous gait analysis technique [17]. Because of the quasi-infinite number of parameters available, it seems reasonable that certain parameters would be best suited for gait analysis in the healthy population.

Systematic reviews have been realized in an attempt to organize and add understanding to the practice of gait analysis in various populations. For example, a systematic review carried out by Sagawa and colleagues [18], using an original methodological approach, was able to identify the most relevant biomechanical parameters for assessing gait in individuals with a lower limb amputation. The results obtained by Sagawa and colleagues [18] leads to question whether the same biomechanical parameters are most relevant for gait analysis in the healthy adult population.

The aim of this systematic review is to determine the most relevant biomechanical parameters used for gait analysis in a healthy adult population. The term relevant was defined as those biomechanical parameters being able to identify gait abnormalities in the healthy adult population and applicable to the clinical and rehabilitation setting. This definition is an adaptation of that used by Sagawa et al. [18].

2. Method

2.1 Procedure for the identification of selected articles

We performed an online search in three databases: PubMed, EMBASE and Web of Science. These three databases were selected for search because of their broad inclusion of multidisciplinary topics within the Biomedical and Health Sciences domain. Each database was searched for all years included in the respective databases with the last search completed in May 2016.

The following search was inputted to all three databases:

> [abstract/title] (Speed OR Cadence OR (Stride time) OR (Swing time) OR (Step time) OR (Single support time) OR (Double support time) OR (Foot flat time) OR (Stance time ratio) OR (Swing time ratio) OR Timing OR (Stride length) OR (Step length) OR (Step width) OR Angle OR Moment OR Power OR (Center of mass) OR (Ground reaction force) OR (Ground reaction impulse) OR (Center of pressure) OR rotation OR symmetry OR velocity OR (stance phase) OR (swing phase) OR (cycle time) OR (spati* temporal) OR hip Or knee OR ankle OR foot) OR (biomechanic*) AND ([MeSH] gait OR walking OR locomotion)

In databases where applicable, certain additional parameters were used to narrow the search. In PubMed, filters including human studies of adults aged 18 to 65 years old, published in French and English and with regards to the nature of the study (i.e. original articles, review articles, case study) were applied. In the EMBASE and Web of Science databases, filters were applied to include human studies, French and English language publications and specific nature of study (i.e. original articles, review articles, case study).

2.2 Inclusion and exclusion criteria

The inclusion and exclusion criteria were developed based upon the purpose of the systematic review, to examine the biomechanical parameters used to study healthy gait. Thus, studies including participants living with pathologies, disabilities, health concerns and/or neurological deficits were excluded. To be selected, articles had to evaluate adults aged 18 to 65 years old with no walking aids. Participants could have been evaluated barefoot, wearing socks, wearing shoes and/or any combination of these three situations. As well, no studies were included if they measured the effect of a treatment or equipment. Selected articles had to at least evaluate participants walking at their self-selected speed on an overground and flat surface.

2.3 Analysis of selected articles

A census of all biomechanical parameters measured was undertaken by two evaluators by carefully reading and analyzing the chosen articles. First, all methodological aspects of the selected articles were tabulated and briefly summarized. Second, the biomechanical parameters measured in all articles were tallied. For each parameter, all articles which measured this parameter were reported and counted. Third, because of the many various instruments, techniques, planes of measurement, etc. used to quantify parameters in the studies selected, the parameters measured were summarized under broader parameter names (i.e.: sagittal, frontal and transverse plane knee power were combined under the broader name of knee power).

Lastly, after a summation of parameters, the number of different articles measuring a type of biomechanical parameter was counted; this was also done for single parameters. Indeed, it seems inevitable to consider not only the most frequently measured parameters, but as well the number of different articles which measure a parameter to observe any disparities between the number of times a parameter was measured versus the number of different articles which measured this parameter.

In an attempt to evaluate relevance of biomechanical parameters, both the frequency of measure and the number of different articles which measure the parameter were combined to produce a score using the summarized parameters. For the first factor, all frequency of measurement scores were divided by the parameter having been measured the most times (hip power: 66 times) and multiplied by 0.5. For the second factor, all number of articles were divided by the parameter having been measured by the most amount of different articles and multiplied by 0.5. Both values were then added to obtain a score weighting both factors. It was deemed that both factors were as important as the other, each contributing to 50% of the score. The following is an example of the calculation for walking velocity, which was measured 50 times by 50 articles:

Walking velocity: ((50/66)*0.5) + ((50/50)*0.5)= 0.879

2.4 Quality of selected articles

We evaluated quality of the selected articles by attributing a Level of Evidence score for each selected article. Our Level of Evidence score was a modified version of that used by Sagawa and colleagues [18], since they were interested in the gait analysis in a population with a lower limb amputation and the current systematic review addresses healthy adult gait analysis. The 14 criteria were subdivided between three main article elements: 1) selection of participants, 2) intervention and assessment, and 3) statistical validity. The maximum possible score was therefore 14, with each article receiving a score of 1 (if they met the requirements) or 0 (if they did not meet the requirements) for each criterion (score of 1 for a non-applicable criterion). Two independent evaluators assessed the score of all articles. For any disparities between scores, both evaluators determined the best suited scoring through discussion. If a consensus could not be reached by the two evaluators, a third evaluator intervened in order to break tie between both scores suggested.

2.5 Data/ Statistical analysis

A Spearman correlation was carried out in order to determine if higher Level of Evidence articles are published in higher Impact Factor journals. Also, a Spearman correlation was sought between the mean Level of Evidence attributed to all articles measuring a given parameter and its frequency of measurement. All statistical analyses were carried out using SPSS 22 (IBM Corp., NY). Level of significance was set at $p < 0.05$.

3.Results:

3.1 Selection of articles

The preliminary database search, using the previously mentioned keyword combination, yielded 16 023 abstracts throughout all three databases. Upon reading the titles and applying the inclusion and exclusion criteria, 1 388 articles were retained for further selection. After reading the abstract, 515 articles remained. Finally, after a careful reading, 65 articles fulfilled the inclusion and exclusion criteria and were selected for further analysis (Figure 1). Table 1 outlines the main methodological aspects of these selected articles.

-------------------------------- Insert figure 1 approximately here ---------------------------

3.2 Participant characteristics

The main participant characteristics of the 65 selected articles are outlined in Table 1.

-------------------------------- Insert Table 1 approximately here ----------------------------

3.3 Article data quality

The Level of Evidence score attributed to each article was in agreement between reviewers. The mean Level of Evidence for all articles was 11.8 ± 1.8, with scores ranging from 6 to 14. The Level of Evidence scores attributed to the 65 articles are outlined in Table 1.

3.4 Parameters for gait analysis

Table 2 indicates that parameters of various types were measured and counted in the selected articles. Parameters from power, work and/or torque were recorded 269 times, spatio-temporal parameters were recorded (256 times), joint angles (177), moments (115) and force (115). A total of 1097 parameters were counted in 65 articles.

All measured biomechanical parameters in the selected articles are outlined in Table 2. The parameter most often measured and/or calculated was the walking speed (50 times) followed by cadence (30 times), stride length (23 times) and step length (21 times).

-------------------------------- Insert Table 2 approximately here ----------------------------

3.5 Parameter summation

As stated, a summation of parameters was carried out (results outlined in Table 3) and the results show that the hip power is the most often measured biomechanical parameter (66 times) followed by the knee power (61 times), walking velocity (50 times) and the ankle angle (47 times).

Also outlined in Table 3 is the number of different articles measuring summarized single parameters. Spatio-temporal parameters were measured in 59 of the 65 articles, angles by 29 different articles and forces in 16 articles. When considering summarized single parameters, walking velocity was measured in 50 different articles and stride length and cadence were measured in 36 and 35 different articles, respectively.

-------------------------------- Insert Table 3 approximately here ----------------------------

The calculation to account for both frequency of measurement and number of articles was carried out with the highest frequency of measurement being the hip power (66 times) and the greatest number of articles being walking velocity (50 articles). Walking velocity obtained the highest score (0.879), followed by stride length (0.686), cadence (0.630), hip power (0.590) and knee power (0.552). The results of this score are presented in Table 4.

-------------------------------- Insert Table 4 approximately here ----------------------------

3.6 Level of Evidence score and Journal Impact Factor

It was sought whether a correlation existed between the article Journal Impact Factor (not shown) and the Level of Evidence score attributed to each article by means of a Spearman correlation. The result of this correlation is a very weak, negative and non-significant correlation (r_s=-0.133, p=0.105). The Impact Factor scores of 4 articles [21,54,59,67] were unavailable and were therefore excluded.

3.7 Frequency of parameters and Level of Evidence score

When the frequency of the most often reported parameters was correlated with the mean Level of Evidence score of articles (not shown), via a Spearman correlation, a weak, negative and non-significant correlation was found (r_s=-0.224, p=0.06).

4. Discussion

4.1 Number of articles

The current review was based on 65 articles. This number may appear small knowing that the review of Sagawa and colleagues [18] included 89 articles of a clinical population. The present study reflects the restrictiveness of our inclusion and exclusion criteria.

4.2 Type, single and summation of biomechanical parameters

4.2.1 Types of biomechanical parameters

Considering types of parameters, it was found that power, work and energy parameters were measured most often (269 times): spatio-temporal parameters followed closely being measured 256 times. Joint angle parameters were measured 177 times, joint moment parameters were measured 115 times and forces were also measured 115 times. In comparison, the systematic review of Sagawa and colleagues [18], revealed that parameters of spatio-temporal type were measured 153 times, joint angles 78 times, platform parameters (i.e. ground reaction forces and center of pressure) 72 times, powers 64 times and joint moments 58 times. Thus, in general, the number of times a type of parameter was measured was less in the review of Sagawa and colleagues [18] than in the present review despite the fact that fewer articles were included for analysis in the current review.

These larger numbers are explained by the fact that both studies did not group parameters in the same manner; therefore, the number of parameters in relation to the total number of articles included in each study is different. Also, Sagawa and colleagues [18] carried out a summation of parameters in which both time sub-parameters and amplitude sub-parameters were grouped separately. For the purpose of our systematic review, it was thought more appropriate to group parameters accordingly, since all are yielded from one measure.

Omitting these disparities, it is possible to note that spatio-temporal parameters are of high relevance in both systematic reviews. As well, all most frequent types of parameters are the same, although they differ in number and order of relevance.

4.2.2 Single parameters

When looking at single parameters, the walking velocity (50 times), cadence (30 times), stride length (23 times) and step length (21 times) were those parameters most frequently measured. These results are in agreement with Sagawa and colleagues [18] who conclude the same parameters were most often measured: walking velocity, cadence, stride and step length. It is interesting to note that for these two different populations the same parameters would appear to be most relevant. This may be because parameters of spatio-temporal type have a certain ease of measurement in comparison to other parameters.

Despite the fact that in the current systematic review, power, work and energy parameters were the most frequently reported measures as a type of parameter, when considering single parameters, the most frequently measured were spatio-temporal. Interestingly, for power, work and energy type of parameters, no single parameter was reported more than 10 times and most parameters were measured only once. In fact, for these types of parameters, a given parameter can be measured at different instances of the gait cycle, in three different planes and for minima and maxima values, making the number of parameters somewhat inflated.

As well, more minima and maxima power values exist at the hip joint when compared to the ankle joint, for example. This may also explain some disparity in the frequency of measurement of some parameters, especially kinematic parameters of the lower limb joints.

4.2.3 Summation of parameters

After a summation of parameters, we observe that those parameters most frequently measured are hip power (66 times), knee power (61 times), walking velocity (50 times) and ankle angle (47 times). Following parameter summation, Sagawa and colleagues [18] concluded walking velocity (43 times), knee angle (31), knee moment (27 times) and hip power (26 times) were most often measured. These differences might reflect that the results are somewhat inflated and the angle, moment and power parameters need to be interpreted cautiously. Indeed, Sagawa and colleagues [18] did not group parameters in the same way as was done in the present review and the frequency obtained for angle, moment and power parameters are smaller. As well, for certain types of parameters (i.e. power, work and energy), the number of total parameters measured (i.e. 269 times) may also be inflated. Again, this may explain some disparity between the number of parameters measured with regards to the total number of articles.

Another explanation for these differences is the type of population studied. Indeed, their choice of clinical population implied the absence of the ankle joint which can explain the lack of ankle joint measures in their population with a transtibial amputation. In the healthy adult population, ankle joint measures were in the top four most relevant parameters after parameter summation.

Also interesting is that articles which measure hip moments, also tend to measure joint moments at the knee and ankle, as they are necessary in inverse dynamic calculations. As well, it is interesting to note that forces are needed in the calculation of moments and angular kinematics are needed for power calculations. Therefore, articles measuring powers, would also measure kinematics, forces and moments and this plays an important role when looking at frequency and relevance of parameters.

In addition to the frequency of measurement, it is also important to consider the number of different articles measuring a given parameter. Out of the total 65 articles included in our systematic review, spatio-temporal parameters were reported by 59 different articles, joint angles were reported by 29 articles, followed by forces (16 articles), joint moments (13 articles) and power, work and energy (13 articles). So for power, work and energy parameters (measured 269 times), the type of parameters which appear to have been measured most often, only 13 out of 65 articles measured these types of parameters. In comparison, spatio-temporal parameters (measured 256 times), were evaluated in 59 of the 65 articles.

As for the type of parameters discussed above, using the summarized parameters, the walking velocity remained the most often measured (50 articles out of 65 total articles) followed by cadence (35 articles), stride length (36 articles), gait cycle parameters (23 articles) and stance time (19 articles). However, when comparing these results to those of Sagawa and colleagues [18], we observe that a higher number of articles reported the most common parameters in our present study: walking velocity was measured only 43 times in 89 articles, cadence 19 times and step and stride length 19 times and 15 times, respectively. An important note must be made here that parameters were summarized differently by both reviews. The differences in the number of articles can, in large part, be explained by the choice of inclusion and exclusion criteria.

As shown by our results, both frequency of measurement and the number of different articles measuring a parameter are of importance when investigating the most relevant biomechanical parameters for gait analysis. The results of the score combining both these factors show that walking velocity, stride length and cadence appear to be most relevant.

4.3 Level of Evidence score

The mean Level of Evidence score for all articles was 11.8 ± 1.8 out of 14 points. This mean score is high; one can argue that it almost reaches a ceiling effect. It is perhaps because the Level of Evidence was not discriminatory enough in the limits for scoring. This can also be due to high quality and soundly based studies. It is perhaps simpler to carry out quality experimentation in a healthy population since there may be less physical restrictions and/or needs as compared to other clinical populations. This may also be due to the inclusion/ exclusion criteria weeding out the lower quality articles. A Level of Evidence score with a wider array of possible scores would be needed.

4.4 Relation between the Level of Evidence score and Journal Impact Factor

The Level of Evidence score of articles were correlated with the Journal Impact Factor. The weak, negative and non-significant Spearman correlation found is in agreement with that of Sagawa and colleagues [18] who carried out this same analysis but with the Journal Impact Factors of the year of publication of their systematic review. It is possible to conclude that both Level of Evidence score and the Journal Impact Factor are not related.

4.5 Relation between the frequency of parameters and their mean Level of Evidence score

The mean Level of Evidence of the articles was correlated with the frequency of the parameter measured. As stated in the results section, a weak, negative and non-significant Spearman correlation was found. It is therefore possible to conclude that the frequency of measurement of a parameter is not related to the mean Level of Evidence of the articles which measure this parameter.

4. 6 Most relevant biomechanical parameters

Spatio-temporal parameters, namely walking velocity, cadence and step and stride length, appear to be the most relevant biomechanical parameters in both individuals with a transtibial amputation and healthy adults. In addition, walking velocity is of even greater relevance since it also measures, and has a direct effect on such parameters as cadence and stride length.

Additionally, these spatio-temporal parameters have a certain ease of measurement: measuring simple spatio-temporal parameters such as walking velocity would appear to be an effective and simple manner to add objectivity to clinical gait analysis which is primarily aimed at ease of measurement [8,13,14].

Future studies should aim to identify if the most relevant biomechanical parameters for gait analysis found in healthy adults are also relevant to other clinical populations. Individuals with a transtibial and transfemoral amputation as well as healthy adults yielded the same most relevant parameters, but perhaps the results obtained in other populations would be different, such as in populations with a neurological disorder (i.e.: Parkinson's, Stroke or Cerebral Palsy) or with a more severe mechanical impairment (i.e.: unilateral hemipelvectomy amputation).

5. Conclusion

A systematic review of the literature pertaining to healthy adult gait was performed and the most relevant biomechanical parameters were identified. Spatio-temporal parameters were those parameters most often measured and by the most amount of articles. Additionally, many specific spatio-temporal parameters were those most often measured (walking velocity, cadence and step/stride length), walking velocity being measured most often, and by the greatest number of articles. Walking velocity, and other spatio-temporal parameters would therefore appear to be the most relevant biomechanical parameters to healthy adult gait analysis.

To our knowledge, this is a first systematic review of its kind in a healthy adult population and the implications of these findings are important for choosing the most relevant biomechanical parameters for gait analysis.

Competing Interests

The authors declare that there are no competing interests.

Acknowledgement

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Reference	Number & Sex of participants	Age of participants	Main Objectives	Level of Evidence
[20]	30 (15 F: 15 M)	20-29	Basic gait data on groups of healthy young adult Kuwaitis of both genders was collected to determine if they duplicated the data published in the Swedish study.	10
[21]	19 (19 M)	25.3 (4.1)	To determine, over two consecutive strides, if the right and left lower limbs developed similar power patterns and if their associated mechanical energies were equal or not in all 3 planes of motion.	14
[22]	17 (8M: 9F)	27.5 (5.3)	To compare overground and treadmill ambulation for possible differences in gait temporal variables and leg joint kinematics.	12
[23]	20 (8 M: 12F)	37-62	To evaluate the time-varying behavior, the test-retest reliability and the concurrent validity of lateral trunk lean and toe-out angles during prolonged walking in healthy adults.	12
[24]	11 (11 M)	28.3 (12.4)	An examination of the angular momenta of healthy adult males walking at three speeds; 0.7, 1.0, and 1.3 times their self-selected comfortable walking speed (CWS).	11
[25]	10 (3F: 7M)	27.5; 20-33	To confirm the hypothesis that stride duration variability exhibits long-range autocorrelations among young healthy subjects walking on level ground, by using an integrated approach that combines distinct methods in order to increase the level of confidence. Also, to determine whether the treadmill disrupts long-range autocorrelations present in stride duration variability and to determine if the outcomes obtained from the treadmill were reproducible across two different testing days.	11
[26]	6 (4M: 2F)	25-45	To improve the understanding of how the central nervous system (CNS) chooses gait parameters for the modulation of velocity by proposing a method for characterizing gait strategies from step frequency and step length analysis.	10
[27]	30 (15F: 15M)	23.6 (2.7)	To define the walking speed and gender effects on the center of pressure (COP) pathway.	11
[28]	98 (51 M: 47 F)	23.5(2.7); 22.9(4.9)	The research hypothesis was that healthy adults would walk differently according to their gender when walking barefoot at their comfortable speed.	14
[29]	14 (8M: 6F)	22.5(3); 23.8(4.1)	To determine if there are changes in temporal gait parameters with a focus on the pelvis when comparing overground and treadmill ambulation, and to assess the effect of sex.	11
[30]	30 (6 groups of 5) (15 M: 15F)	20-30; 31-45; 46-60	To investigate the effects of age, gender and walking speed on different gait performance measures including joint motion, ground reaction forces (GRF), electromyography (EMG), heart rate (HR), and perceived exertion during walking at different percentage of preferred walking speed (PPWS).	12
[31]	8 (6M: 2F)	22-30	To examine trunk, neck and head movements to determine a mechanism for upper body stabilization during walking.	12
[32]	10 (5M: 5F)	27.10 (3.25)	To demonstrate that the processes responsible for maintaining local dynamic stability of walking act across multiple consecutive strides of gait.	11
[33]	14 (4M: 10F)	30-55	To analyze foot and ankle kinematics from gait recordings of healthy subjects walking at comfortable and slower speeds.	11
[34]	10 (7M: 3F)	23 (2)	To analyze the 3D angle between the joint moment and the joint angular velocity vectors at the ankle, knee and hip during the gait cycle and to investigate if these joints are predominantly driven or stabilized.	11
[35]	46 (32M: 14F)	--	Velocity, stride length and stride frequency were treated as independent variables in relation to each other in a graphic form to see how they interact in gait. To achieve this, a Velocity Field Diagram (VFD) was described.	6
[36]	9 (9M)	28.5 (5)	To characterize the basic features of the moment-angle curves in normal walking at different velocities.	12
[37]	39 (21M: 18F)	27(4.2); 22.9 (4.1)	To characterize and compare the dynamic joint stiffness (DJS) of the ankle in the sagittal plane during natural cadence walking in both genders.	12
[38]	10 (10M)	23.3 (2.4)	To investigate the variability and symmetry of ground reaction force (GRF) measurements during walking, using time and frequency domain analysis.	13
[39]	16 (16M)	22.8 (1.6)	To demonstrate that data from a video-based system could be used to estimate the net effect of the external forces during gait, to determine the contribution of the trunk and upper and lower limbs using their accelerated body masses, and to test the hypothesis that the thigh mainly assumed lower limb propulsion during able-bodied locomotion.	14
[40]	20 (20M)	23.8 (2.2.)	To investigate the changes in horizontal velocity which are known to influence many biomechanical characteristics of human locomotion, with respect to the interlimb symmetry of walking in a normal population.	14
[41]	14 (8M: 6F)	19-56	1) To determine whether asymmetries exist between limbs of healthy individuals during gait and 2) to examine the relationship between lower extremity lateral dominance and any observed differences.	12
[42]	10 (10M)	18-29	1) To determine whether long-range correlations in gait extend over very long-time scales; 2) to define the conditions under which such correlations may exist; and 3) to evaluate potential mechanisms underlying this fractal property of gait.	11
[43]	11(11F)	27.4(4.0); 22-30	To investigate the influence of walking speed on the amount and structure of the stride-to-stride fluctuations of the gait cycle.	9
[44]	13 (7M: 6F)	23.3(3.0)	To examine how gait speed influences healthy individual's lower trunk motion during overground walking and to assess if Principal Component Analysis (PCA) can be used to gain further insight into postural responses that occur at different walking speeds.	13
[45]	10	23 (4)	To investigate the relationship between oscillatory dynamics of the head and trunk in each plane of motion during walking.	13
[46]	68 (32M: 36F)	34 (11)	To examine the changes, if any that occur in peak lower extremity net joint moments while walking in industry recommended athletic footwear.	12
[47]	110 (57F: 53M)	29.1(8.9); 28.3(5.04)	To determine if knee joint torques, which are likely relevant to the development and, possibly, progression of knee osteoarthritis, are equivalent between genders during natural, barefoot walking.	14
[48]	30 (17M: 13F)	24.6 (4.0)	To evaluate the effect of pelvic rotation, originally described as the first determinant of gait, on reducing the vertical displacement of the center of mass (COM) during comfortable speed walking.	13

[49]	20 (10M: 10F)	27-56	To determine three-dimensional foot and ankle kinematics, using a three-segment foot model and to determine ground reaction forces, temporal force factors and time-related factors in normal subjects.	12
[50]	25 (25M)	26.2 (5.2)	To test if the lower limb joint and thoraco-lumbar moments are similar in subjects who maintain an average natural forward or backward trunk inclination during gait and verify if the lower limbs are equally affected.	12
[51]	16 (8M: 8F)	18-28	To study the familiarization time required for reliable sagittal-plane knee kinematics and temporal-distance gait measurements to be obtained from treadmill walking and whether knee kinematics and temporal-distance gait measurements obtained from familiarized treadmill walking can be generalized to overground walking.	14
[52]	20 (5M: 15F)	18-30	To investigate the short-term relationships between footstep variables during steady state, straight-line, over-ground walking in healthy adults and to explore the extent to which the performance of a step or stride is dependent on the performance of an earlier step or stride in a sequence.	12
[53]	10 (7M: 3F)	26.9 (5.7)	To examine the effect of walking speed on center of mass (COM) displacement in the medial-lateral (ML) and vertical directions.	13
[54]	10 (6M: 4F)	23 (5)	1) Quantifying gait pseudo-periodicity using information concerning a single stride; 2) investigating the effects of walking pathway length on gait periodicity; 3) investigating separately the periodicity of the upper and lower body part movements; 4) verifying the validity of foot-floor contact events as markers of the gait cycle period.	12
[55]	8 (4M: 4F)	24-38	To determine if walking at the predicted frequency produced greater shock attenuation through the body when compared with other frequencies at the same walking speed and to assess the role played by the individual segments in attenuating shock under different frequency-stride length combinations at a constant speed.	12
[56]	26 (13M: 13F)	18-35	(1) To compare the kinematics of treadmill gait to overground gait obtained in laboratory, comparing the present findings to those previously reported and (2) to quantify any kinetic differences between overground and treadmill gait, including, for the first time an analysis of the joint moments and powers of treadmill gait.	7
[57]	48 (10M: 38F)	23-62	To simultaneously statistically test whether the three factors gender, age and walking speed significantly affect kinematic gait data in a reference population.	14
[58]	22 (9M: 13F)	35-55	To determine if the variability in the characteristics of the net external hip adduction moment can be explained by the strength of the hip abductor musculature, subject anthropometrics, gait velocity and the corresponding characteristics of the gluteus medius electromyogram captured during gait in healthy individuals.	12
[59]	32 (20M: 12 F)	24.9 (2): 24.1 (1.6)	Gait analysis was conducted on Korean subjects in their 20s and these gait characteristics were compared to those reported in previously published studies conducted in Western countries.	13
[60]	20 (20M)	25.3 (4.1)	To test the hypothesis that limb propulsion is mainly associated with the interaction of a number of muscle power bursts developed throughout the stance phase and that the control actions are mainly achieved by the contralateral limb through different power-burst interactions.	12
[61]	19 (19M)	26.2 (3.2)	To test the hypothesis that the trailing limb contributes mainly to forward progression, whereas the trailing limb provides control and propels the lower limb to a lesser extent.	14
[62]	20 (20 M)	25.3 (4.1)	(a) To identify the main functions of the ankle and hip muscle moments and their contribution to support and propulsion tasks, and (b) to illustrate the interaction between ankle and hip moment activities.	14
[63]	19 (19 M)	25.3 (4.1)	To demonstrate that the ankle frontal muscle power absorption and generation at push-off are related to the foot's initial position at heel-strike with respect to the body center of mass.	13
[64]	20 (10M: 10F)	24 (3)	To compare bilateral ground reaction force impulses to evaluate functional asymmetry as an explanation for gait asymmetries.	13
[65]	25 (8M: 17 F)	19-32	To report the reproducibility of the invariant walk ratio in repeated trials involving young healthy adults walking at a variety of speeds.	12
[66]	22 (10M: 12F)	25.9 (4.1): 20.6 (1.4)	To examine whether there is an optimal walking speed with minimum intrasubject variability in step length and step width during free walk and whether there is an optimal step rate with minimum step length variability during walking with imposed step rates.	12
[67]	28 (14M: 14F)	20-34	To test the applicability a control scheme to the unconstrained portion of the gait cycle- the swing phase.	11
[68]	40 (20M: 20F)	24.1 (3.1): 22.5 (3.2)	To determine the kinematic variability of the lower extremity joints using methods from the mathematical chaos theory in a normal walking environment in conjunction with a large population of healthy young adults and to test the hypothesis that variability characteristics are different between joints and to further investigate differences between male and female and right and left subgroups.	13
[69]	10 (5M: 5F)	19-34	1) To introduce the knee moment arm length as a measure to evaluate knee pre- and postoperatively; (2) to determine the variability in trials done minutes apart and trials done days apart; (3) to present some normative data for healthy subjects for use as reference values in assessment of patients with knee deformities; and (4) to determine the variability in the hip, knee and ankle moments in the frontal and sagittal planes, in trials done minutes apart and days apart.	11
[70]	16 (slow:3M: 5F) (fast: 3M: 5F)	Slow: 20.74 Fast: 19.75	To determine the familiarization period required to obtain consistent measurements of the angular movements of the lumbar spine and pelvis during treadmill walking.	13
[71]	27 (slow: 7M: 6F)	Slow: 23.5 (5.1)	To study the effect of walking at a self-selected and at a slower speed on the angular movements of the pelvis and lumbar spine and how interpretation of	8

	(fast: 5M: 9F)	Fast: 20.6 (2.8)	speed effects on lumbar spine movements was influenced by frame of reference, either relative to the pelvis or relative to a global reference frame.	
[72]	14 (7M: 7F)	46 (13.3)	To employ an analytical model to estimate the effects of walking cadence and laterality on the positive and negative mechanical work performed by the hip, knee and ankle muscles in the sagittal plane.	12
[73]	8 (3M: 5F)	23-34	To measure the mechanical energy changes of the center of gravity (CG) of the body in forward, lateral and vertical direction during normal level walking at intermediate and low speeds.	11
[74]	18 (9M: 9F)	35.9 (10)	To test the 2D PL (power law) compliance of motion of the center of mass (CM) within the step, as a premise to further 3D modeling, so far applied to upper limb motion.	11
[75]	62 (21M: 41 F)	41.4 (11.0)	To investigate if the detailed pressure data of the footprints of normal gait add essential information to the spatio-temporal variables of gait.	6
[76]	19 (19M)	25.3 (4.1)	To determine if more than one gait pattern exists in able-bodied young men, by analyzing the dissimilarities in the three-dimensional (3-D) muscle powers developed at the joints of the right lower limb.	14
[77]	9 (9M)	28.7 (4.4)	To determine the differences between angular oscillation curves of the lumbar spine and pelvis during walkway and treadmill ambulation.	14
[78]	15 (4M: 11F)	25.5 (4.5)	To determine if limb dominance affects the vertical ground reaction force and center of pressure (COP) during able-bodied gait.	9
[79]	10 (5M: 5F)	24.3 (4.0)	Sole-floor reaction forces were measured from five anatomically discrete points in the human sole during locomotion on the treadmill and on the laboratory floor.	14
[80]	24 (11M: 13 F)	27 (7)	To compare vertical ground reaction forces walking overground with vertical foot-belt forces for treadmill gait.	10
[81]	20 (9M: 11F)	24 (4)	To investigate the contribution of passive mechanisms to lower extremity joint kinetics in normal walking at slow, comfortable and fast speeds.	12
[82]	12	28.5 (3.3)	To investigate whether multiple short bouts of gait can be used for the valid and reliable assessment of variability and local dynamic stability, and how many bouts are required for their reliable estimation.	11
[83]	21 (10M: 11F)	26.9 (4.5)	To assess the validity of the anatomical landmark data derived from the Kinect's skeleton tracking algorithm for examining the spatiotemporal characteristics of gait in young, healthy individuals.	11
[84]	10 (10M)	28.8 (8.3)	To demonstrate how vector field statistics can be used to more objectively analyse CoP trajectories.	10

Table 1- Methodological aspects of selected articles.[1]

POWER, WORK & TORQUE (**269**)					
Parameter	Total	Articles	Parameter	Total	Articles
Sagittal hip power peak 1	5	[60], [61], [72], [76], [81]	**Sagittal hip power peak 2**	2	[60], [61]
Sagittal hip power peak 2	5	[60], [61], [72], [76], [81]	**Sagittal hip power peak 3**	2	[60], [61]
Sagittal hip power peak 3	5	[60], [61], [72], [76], [81]	**Frontal hip power peak 1**	2	[60], [61]
Sagittal knee power peak 1	5	[60], [61], [72], [76], [81]	**Frontal hip power peak 2**	2	[60], [61]
Sagittal knee power peak 2	5	[60], [61], [72], [76], [81]	**Frontal hip power peak 4**	2	[60], [61]
Sagittal knee power peak 3	5	[60], [61], [72], [76], [81]	**Transverse hip energy peak 1**	2	[60], [61]
Sagittal ankle power peak 1	5	[60], [61], [72], [76], [81]	**Transverse hip energy peak 2**	2	[60], [61]
Sagittal ankle power peak 2	5	[60], [61], [72], [76], [81]	**Transverse hip energy peak 3**	2	[60], [61]
Frontal ankle power peak 2	4	[60], [61], [76], [63]	**Sagittal knee energy peak 1**	2	[60], [61]
Frontal hip power peak 1	3	[60], [61], [76]	**Sagittal knee energy peak 2**	2	[60], [61]
Frontal hip power peak 2	3	[60], [61], [76]	**Sagittal knee energy peak 3**	2	[60], [61]
Frontal hip power peak 3	3	[60], [61], [76]	**Frontal knee energy peak 1**	2	[60], [61]
Transverse hip power peak 1	3	[60], [61], [76]	**Frontal knee energy peak 2**	2	[60], [61]
Transverse hip power peak 2	3	[60], [61], [76]	**Transverse knee energy peak 1**	2	[60], [61]
Transverse hip power peak 3	3	[60], [61], [76]	**Transverse knee energy peak 2**	2	[60], [61]

[1] Methodological aspects of all selected articles. The above chart depicts the reference, number of participants, sex of participants (M for male: F for female), the participant age (mean, standard deviation in parentheses and range separated by a hyphen), as well as the main objectives of the study and the Level of Evidence score attributed to each article (on a possible 14 total points). All available information concerning participant characteristics was provided. If the articles reported participant age and sex characteristics per group (i.e. fast and slow walking group), the information is provided as such.

Sagittal knee power peak 4	3	[72], [76], [81]	**Transverse knee energy peak 3**	2	[60], [61]
Frontal knee power peak 1	3	[60], [61], [76]	**Sagittal ankle energy peak 1**	2	[60], [61]
Frontal knee power peak 2	3	[60], [61], [76]	**Sagittal ankle energy peak 2**	2	[60], [61]
Transverse knee power peak 1	3	[60], [61], [76]	**Frontal ankle energy peak 1**	2	[60], [61]
Transverse knee power peak 2	3	[60], [61], [76]	**Frontal ankle energy peak 2**	2	[60], [61]
Transverse knee power peak 3	3	[60], [61], [76]	**Sagittal plane knee power**	2	[21], [81]
Frontal ankle power peak 1	3	[60], [61], [76]	**Sagittal plane hip power**	2	[21], [81]
Frontal hip power peak 3	3	[60], [61], [63]	**Sagittal plane ankle power**	2	[21], [81]
Frontal hip power peak 4	2	[60], [61]	**Frontal plane ankle power**	2	[21], [63]
Sagittal hip power peak 1	2	[60], [61]			
SPATIO-TEMPORAL PARAMETERS **(256)**					
Parameter	**Total**	**Articles**	**Parameter**	**Total**	**Articles**
Walking velocity	50	[20], [76], [60], [61], [60],[61],[76], [21], [81], [27], [28], [29], [30], [32], [33], [40], [42], [43], [44], [45], [46], [48], [50], [51], [52], [53], [55], [56], [57], [58], [59], [60], [61], [72], [73], [74], [75], [76], [77], [79], [81], [63], [64], [65], [66], [67], [71], [80], [83], [84]	**Stride width**	4	[28], [33], [53], [59]
Cadence	30	[20], [22], [21], [37], [24], [26], [27], [28], [29], [30], [49], [50], [51], [52], [53], [54], [55], [56], [59], [60], [61], [72], [75], [76], [79], [63], [65], [66], [67], [80]	**Swing time**	4	[22], [24], [59], [75]
Stride length	23	[22], [21], [35], [36], [24], [28], [32], [33], [43], [50], [51], [53], [56], [59], [60], [61], [72], [75], [76], [63], [67], [80], [83]	**% Stance time**	3	[21], [38], [28]
Step length	21	[20], [21], [26], [33], [41], [43], [44], [48], [52], [53], [54], [59], [60], [61], [73], [75], [76], [65], [66], [67], [83]	**Step time**	4	[20], [41], [75], [83]
Stance time	12	[38], [24], [29], [39], [40], [41], [49], [59], [60], [61], [75], [79]	**Time of heelstrike**	3	[25], [31], [39]
Gait cycle (%)	10	[34], [28], [29], [44], [46], [47], [50], [54], [57], [59]	**Time of toe-off**	3	[31], [33], [39]
Stride time	12	[25], [22], [35], [29], [32], [33], [44], [51], [54], [56], [80], [83]	**Breaking phase**	2	[60], [61]
Double support time	7	[24], [31], [33], [52], [60], [61], [75]	**Stride interval**	2	[42], [43]
% Stance phase	6	[33], [49], [50], [75], [76], [63]	**Stride frequency**	2	[35], [32]
Gait cycle time	5	[41], [49], [75], [77], [67]	**Terminal double support time**	2	[60], [61]
% Double support	4	[21], [28], [56], [75]	**Step width**	2	[54], [66]
ANGLES **(177)**					
Parameter	**Total**	**Articles**	**Parameter**	**Total**	**Articles**

Maximum ankle sagittal plane dorsiflexion	4	[30], [41], [49], [56]	**Maximum hip sagittal plane flexion**	2	[30], [56]
Maximum ankle sagittal plane plantarflexion	4	[30], [41], [49], [56]	**Maximum hip sagittal plane extension**	2	[30], [56]
Hip sagittal angle	3	[22], [81], [67]	**Maximum knee sagittal plane extension**	2	[41], [56]
Knee sagittal angle	3	[22], [81], [67]	**Frontal plane ankle angular velocity**	2	[21], [63]
Ankle sagittal angle	3	[22], [81], [67]	**Upward maximum pelvic obliquity angle**	2	[29], [56]
Maximum knee sagittal plane flexion	3	[30], [41], [56]	**Downward minimum pelvic obliquity angle**	2	[29], [56]
Foot progression angle	2	[57], [58]	**Max knee extension angle**	2	[22], [32]
Sagittal plane ankle angle position	2	[36], [49]	**Pelvic rotation angle**	2	[48], [57]
Sagittal plane hip angle position	2	[36], [28]			
MOMENTS **(115)**					
Parameter	**Total**	**Articles**	**Parameter**	**Total**	**Articles**
Sagittal plane hip moment	6	[21], [36], [46], [50], [81], [60]	**Peak knee flexion moment**	2	[46], [56]
Sagittal plane ankle moment	6	[21], [36], [46], [50], [81], [60]	**Peak knee varus moment 1**	2	[46], [56]
Sagittal plane knee moment	5	[21], [36], [46], [50], [81]	**Peak knee external rotation moment**	2	[46], [56]
Peak hip extension moment	4	[46], [50], [56], [60]	**Peak knee internal rotation moment**	2	[46], [56]
Peak hip flexion moment	4	[46], [50], [56], [60]	**Peak ankle eversion moment**	2	[46], [56]
Peak ankle dorsiflexion moment	3	[46], [56], [60]	**Peak ankle external rotation moment**	2	[46], [56]
Peak ankle plantarflexion moment	3	[50], [56], [60]	**Peak ankle internal rotation moment**	2	[46], [56]
Frontal plane ankle moment	3	[21], [46], [63]	**Transverse plane knee moment**	2	[21], [46]
Peak knee extension moment 1	2	[50], [56]	**Frontal plane knee moment**	2	[21], [46]
Peak hip adduction moment 1	2	[46], [56]	**Transverse plane hip moment**	2	[21], [46]
Peak hip external rotation moment	2	[46], [56]	**Frontal plane hip moment**	2	[21], [46]
Peak hip internal rotation moment	2	[46], [56]	**Transverse plane ankle moment**	2	[21], [46]
FORCES **(115)**					
Parameter	**Total**	**Articles**	**Parameter**	**Total**	**Articles**
Fz1	8	[38], [30], [40], [43], [46], [49], [78], [80]	**V1 (vertical maximum force)**	3	[39], [56], [59]
Fz3	8	[38], [30], [40], [43], [46], [49], [78], [80]	**S1 (sagittal maximum force)**	3	[39], [56], [59]
Fz2	7	[38], [30], [40], [43], [49], [78], [80]	**T1 (maximum transverse force)**	3	[39], [56], [59]
Fy1	4	[38], [40], [46], [49]	**Fx3**	2	[46], [49]
Fy2	4	[38], [40], [46], [49]	**Time to Fx1**	2	[38], [49]
Fx1	3	[38], [46], [49]	**Time to Fx2**	2	[38], [49]
Fx2	3	[38], [46], [49]	**V3 (vertical maximum force)**	2	[39], [59]
Time to Fz1	4	[38], [40], [49], [80]	**S2 (sagittal minimum force)**	2	[39], [56]
Time to Fz2	4	[38], [40], [49], [80]	**S3 (sagittal maximum force)**	2	[39], [59]
Time to Fz3	4	[38], [40], [49], [80]	**T2 (minimum transverse force)**	2	[39], [56]
Time to Fy1	3	[38], [40], [49]	**T4 (maximum transverse force)**	2	[39], [59]
Time to Fy2	3	[38], [40], [49]	**Ground reaction forces AP**	2	[74], [64]
ACCELLERATION **(52)**					
Parameter	**Total**	**Articles**	**Parameter**	**Total**	**Articles**

VT head acceleration	2	[45], [55]	**AP head acceleration**	2	[45], [55]
SYMMETRY **(25)**					
Parameter	**Total**	**Articles**	**Parameter**	**Total**	**Articles**
Symmetry of Fz1	2	[38], [78]	**Symmetry of Fz3**	2	[38], [78]
Symmetry of Fz2	2	[38], [78]	**AP COP displacement**	2	[78]
CENTER OF MASS **(22)**					
Parameter	**Total**	**Articles**	**Parameter**	**Total**	**Articles**
COM displacement (vertical)	3	[48], [53], [74]	**COM velocity (VT)**	2	[53], [74]
COM displacement (M/L)	2	[53], [74]	**COM velocity (M/L)**	2	[53], [74]
COM displacement (A/P)	2	[74], [63]			
LOCAL DYNAMIC STABILITY **(19)**					
VARIABILITY **(18)**					
COP **(11)**					
OTHER (36)					

Table 2- Measured biomechanical parameters.[2]

POWER, WORK & TORQUE **(269)** 13					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
Hip power	66	9	**Limb energy**	6	1
Knee power	61	9	**Limb work**	6	1
Ankle power	36	9	**Foot momentum**	6	1
Arm momentum	12	1	**Shank momentum**	6	1
Energy	8	1	**Thigh momentum**	6	1
Hip work	6	1	**Knee torque**	5	2
Knee work	6	1	**Head & neck momentum**	3	1
Ankle work	6	1	**Torso momentum**	3	1
Hip energy	6	1	**Total body momentum**	3	1
Knee energy	6	1	**Hip torque**	1	1
Ankle power	6	1	**Ankle torque**	1	1
SPATIO-TEMPORAL PARAMETERS **(256)** 59					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
Walking velocity	50	50	**Double support**	13	11
Stride length	43	36	**Step width**	6	6
Gait cycle	37	23	**Swing time**	6	5
Cadence	37	35	**Single support**	2	2
Stance time	22	19	**OTHER**	14	4
Stride time	16	16			
ANGLES **(177)** 29					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
Ankle angle	47	17	**Lumbar angle**	7	1
Pelvis angle	37	9	**Spine angle**	6	1
Hip angle	30	13	**Neck angle**	2	1
Knee angle	29	14	**Head angle**	2	1
Trunk angle	8	4	**Sacrum angle**	1	1
Thorax angle	7	1			
MOMENTS **(115)** 13					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
L5 moment	5	1	**Ankle moment**	35	12
Hip moment	37	11	**Other**	8	2
Knee moment	30	9			
FORCES **(115)** 16					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
Vertical ground reaction force	43	13	**Lower limb ground reaction forces**	3	1
Anterior-posterior ground reaction forces	27	9	**Upper limb ground reaction forces**	3	1

[2] Biomechanical parameters measured in included studies. This chart tabulates each biomechanical parameter as it was measured in the designated study. The reference measuring each given parameter is given, as well as the total for single parameters. The following parameters are grouped according to their type and a total of number of parameters measured per type is given in parentheses. Only parameters measured more than once are shown here.

Medial-lateral ground reaction forces	22	6	**Other**	8	3
Head and trunk ground reaction forces	3	1			
ACCELLERATION **(52)** 2					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
Head velocity	21	2	**Ankle velocity**	6	1
Trunk velocity	13	1	**Knee velocity**	4	1
Shoulder velocity	6	1	**Ankle velocity**	2	1
SYMMETRY **(25)** 4					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
Ground reaction symmetry	12	3	**COP symmetry**	7	2
Spatio-temporal symmetry	8	1			
CENTER OF MASS & GRAVITY **(22)** 4					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
Center of mass	19	4	**Center of gravity**	3	1
LOCAL DYNAMIC STABILITY **(19)** 3					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
Local Dynamic Stability	18	3			
VARIABILITY **(18)** 3					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
Coefficient of variation	8	1	**Standard deviation**	7	1
COP **(11)** 3					
Parameter	**Total**	**Number of articles**	**Parameter**	**Total**	**Number of articles**
COP velocity	5	2	**COP position**	2	1
COP displacement	3	1			
OTHER **(36)** 7					

Table 3- Summation of parameters.[3]

Parameter	Frequency	Number of articles	Relevance score
POWER, WORK & TORQUE			
Hip power	66	9	0.590
Knee power	61	9	0.552
Ankle power	36	9	0.363
SPATIO-TEMPORAL PARAMETERS			
Walking velocity	50	50	0.879
Stride length	43	36	0.686
Cadence	37	35	0.510
Gait cycle	37	23	0.630
Stance time	22	19	0.357
ANGLES			
Ankle angle	47	17	0.526
Pelvis angle	37	9	0.370
Knee angle	29	14	0.360
Hip angle	30	13	0.357
FORCES			
Vertical ground reaction force	43	13	0.456

[3] Summation of all biomechanical parameters measured in included studies. This chart tabulates each parameter under a broader theme of parameters as well as the number of different articles which measure this summarized parameter. The total number of parameters measured per type is shown in parentheses beside the parameter type; the total number of different articles measuring a type of parameter is given beside these parentheses. The breakdown of the summation is not shown here. Only parameters measured more than once are shown.

Table 4- Relevance score. [4]

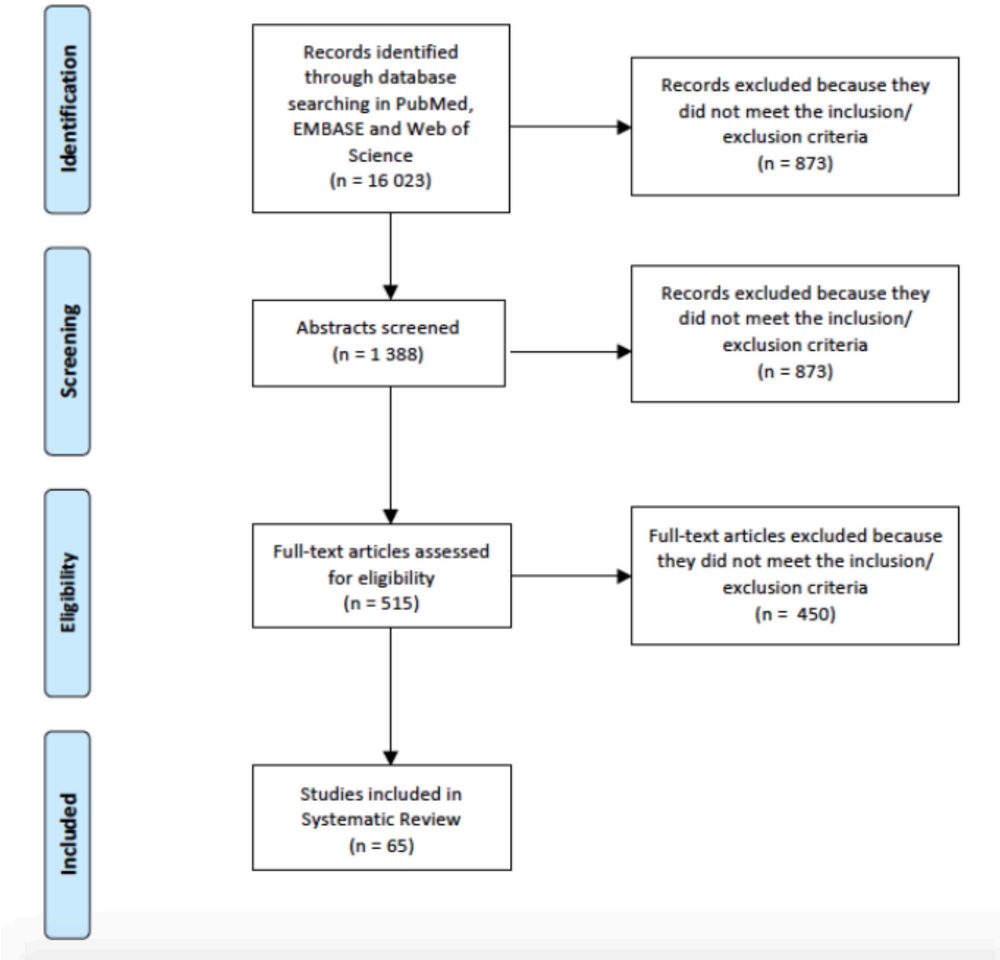

Figure 1- Article selection flowchart
Flowchart as per PRISMA guidelines (19) summarizing the procedure for the selection of articles after the interrogation of three databases. All articles were retained or dismissed for analysis by the application of the inclusion and exclusion criteria (see methods). First, the articles were retained or dismissed on the basis of the article titles. A second step consisted of the reading of the article abstracts. Finally, all retained articles were read and a final selection was made.

[4] This relevance score is calculated based on the frequency of measurement and the number of different articles measuring the given parameter, as described in the methods section. Only parameters which scored more than 0.300 are shown here.

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Chapter 4: Article II

Anticipatory postural adjustments in the dysvascular transtibial amputee during gait initiation.

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Abstract

In gait initiation, anticipatory postural adjustments (APA) allow for total center of mass to be transferred from bipedal to single limb stance, enabling step initiation and propulsion while maintaining balance. In healthy adults, these APA take place in both antero-posterior (A/P) and medio-lateral (M/L) directions and follow sequential phases. In the unilateral transtibial amputee (TTA), studies investigating gait initiation APA have been fewer, and though the dysvascular TTA (DTTA) population is the most sizeable and growing, no studies have yet investigated the APA gait initiation pattern in this specific population of DTTA. Thus, the purpose of the current study was to characterize the APA strategy employed by DTTA when compared to their age-matched controls. On a walkway embedded with three force plates, standing with both feet on either side-by-side force plates, 10 DTTA and 10 control participants were asked to self-initiate gait, stepping onto the third force plate, completing five trials with each limb leading. Parameters for phases 1-3 and total APA net center of pressure displacement (CoP_{net}) were calculated in A/P and M/L directions. A significant reduction in phase 3 M/L CoP_{net} displacement was observed in the intact limb (3.59±0.29 cm) when compared to control and prosthetic limbs (7.03±2.00 and 7.11±0.40 cm, respectively; p=0.05 and p=0.04, respectively) which can be explained by limb load asymmetry observed in the DTTA. As well, an anterior total CoP_{net} displacement was observed beneath the prosthetic limb (3.20±2.96 cm), an important result in the DTTA. Previous studies have found both anterior and reduced posterior total CoP_{net} displacement, prior studies investigating APA in both traumatic and dysvascular TTA confounded. However, when compared to the TTA, the DTTA are deconditioned and inactive. The total anterior CoP_{net} displacement observed would appear to be related to

further reductions in stability caused by reduced fitness levels, sensory loss, peripheral dysvascularity in the intact limb, etc. associated with dysvascular amputation. The APA strategy utilized by the DTTA favor stability over propulsion.

1. Introduction

Anticipatory postural adjustments (APA) are a feedforward control process in preparation and planning of a movement. APA allow for total center of mass (CoM) to be transferred from a position of bipedal to single limb stance, to enable movement initiation and propulsion while maintaining equilibrium [1]. The net center of pressure (CoP_{net}) and CoM have been proposed to interact as an inverted pendulum, the CoP_{net} acting as the independent variable, pushing and pulling the total CoM and maintaining balance throughout quiet standing, gait initiation, steady-state walking and termination [2]. APA are a result of preparation to voluntary movement and are present prior to the onset of movement [3]. APA are pre-programmed and specific to the voluntary movement wanted. In gait initiation, APA follow a sequential and determined pattern [4].

Gait initiation is a complex motor task, requiring important coordination and balance of the body to transfer body weight to one limb as the other limb is propulsed forward into the first step while resisting gravity [5,6]. To create gait initiation, proper APA must take place [7,8].

In the healthy adult, APA prior to gait initiation have been well documented [9-12]. Propulsion during gait initiation is achieved by the APA prior to the first step with the leading limb, and then the APA beneath the trailing limb produce further propulsion during gait initiation [8,13]. These APA take place in both the antero-posterior (A/P) and medio-lateral (M/L) directions and are described by the following sequential phases: 1) a displacement of the CoP_{net} posteriorly and laterally toward the leading limb (Figure 1- APA1); 2) an anterior displacement of the CoP_{net} towards the trailing limb, as weight is loaded to the trailing limb, and the end of the second phase occurs when the CoP_{net} is approximately centered between both limbs and there is leading limb heel-off (Figure 1- APA2); 3) A posterior and lateral displacement towards and beneath the trailing limb takes place with leading limb toe-off (Figure 1- APA3); and finally, 4) there is a rapid displacement of the CoP_{net} with the trailing limb toe-off, the CoP_{net} travelling from heel to toe-off (Figure 1-APA4). Indeed, the APA's are referred to as anticipatory (i.e. prior to gait initiation) but they indeed continue through to toe-off of the trailing limb [14].

---------------------------------------Insert Figure 1 about here-------------------------------

In the unilateral transtibial amputee (TTA) without distinction for cause of amputation (i.e. traumatic, dysvascular, etc.), studies investigating APA during gait initiation have been fewer. In the TTA, initiating gait is preferred with the amputated limb (i.e. as leading limb) as in this situation, body weight is loaded to the intact (trailing) limb, a more stable situation [15,16]. As well, the TTA favour stability over propulsion, causing a reduced speed in gait initiation, partially the result of loss of musculature and sensory information from the amputated limb [17]. Thus, the APA's are therefore also impacted.

When gait is initiated with the amputated limb, the posterior CoP_{net} displacement beneath the intact trailing limb is diminished (i.e. APA3 phase) [8,16,17,18]. When gait is initiated with the intact limb, the CoP_{net} displacement beneath the amputated trailing limb is even further reduced, some results even providing evidence of a slight anterior, rather than posterior, shift in the CoP_{net} [8,16,17,18]. The path of the CoP_{net} displacement remains the same as in healthy controls when gait is initiated with the prosthetic limb, but differs when gait is initiated with the intact limb, the non-preferred leading limb. However, the CoP_{net} trajectory is altered beneath the trailing prosthetic limb [8,15,16].

Finally, an important APA change to gait initiation in the TTA when compared to healthy adults is the increase in time needed to complete the task, the main time increase occurring during the posterior APA3 phase beneath the trailing limb [16,25].

Since CoP_{net} and CoM act together as an inverted pendulum to propel the body forward into gait, a reduction in a posterior CoP_{net} displacement creates a reduced torque on the CoM, and consequently, a reduced forward propulsion of the total body. Further still, if the CoP_{net} displacement is anterior, torque generation is further reduced in producing forward propulsion (i.e. as step has been initiated in the leading limb, the CoM remains anterior when compared to the CoP_{net}). By increasing the time taken to complete the APA in gait initiation, it is thought that the TTA somewhat counters the lack of propulsion created by the reduced inverted pendulum unbalance. Though posterior CoP_{net} displacement is reduced, the TTA are thus able to create some forward propulsion [8,15,16].

Of particular interest is the sizeable and growing population of dysvascular TTA (DTTA). Type II diabetes, the main cause for DTTA, is projected to increase to 360 million people by year 2030 and the number of dysvascular amputations is projected to double by year 2050 [19]. Moreover, the DTTA is the most important major amputation, (i.e. omitting toe and finger amputation) [20]. Studies have explored the important and specific constraints posed in the DTTA when compared to their traumatic counterparts as confounding health factors are often present in the DTTA, posing additional challenges to amputation, such as peripheral dysvascularity in the non-amputated leg, sensory and vision loss, important physical deconditioning, etc. [21,22]. Consequently, survival rate is lower in the DTTA population [23]. Reduced steady-state walking velocity, increased sway in quiet standing and inability to balance on one leg are but some of the principal differences that have been observed in the DTTA when compared to traumatic TTA [21,24,25].

It is therefore plausible that strategies employed in APA prior and during gait initiation would also be unique in the DTTA yet no studies have explored the APA pattern prior to gait initiation in the DTTA population. Thus, the purpose of the current study is therefore to compare the APA pattern, both CoP_{net} displacement and time, employed by DTTA when compared to their age-matched controls.

2. Methods

2.1. Subjects

Ten unilateral DTTA participants were recruited via the Institut de réadaptation Gingras-Lindsay de Montréal and ten control participants were recruited via acquaintances of the researchers at Université de Montréal. Control participants were healthy individuals age-matched to the DTTA subjects. For all participants, any disease or condition having an impact on the locomotor or postural pattern (other than cause of dysvascular amputation in DTTA group- i.e. Type II Diabetes) was reason for participant exclusion. All DTTA wore their own passive prosthesis, having been ambulant for at least 12 months prior to testing with their prosthesis. Participants provided informed consent prior to testing and participant characteristics are summarized in Table 1.

---------------------------------Insert Table 1 approximately here----------------------------

This study was approved by the Comité d'éthique de la recherché en santé de l'Université de Montréal and the Comité d'éthique en recherche des établissements du Centre de recherche interdisciplinaire de réadaptation du Montréal métropolitain.

2.2. Equipment

A walkway with three AccuGait AMTI (Advanced Medical Technology Inc., MA) embedded force plates was set up as is displayed in Figure 1. The force plates measured forces and moments and was sampled at 100 Hz. Data analysis was carried out using a MATLAB program (The MathWorks Inc., MA) created for the purpose of the present research project.

2.3.Procedure

Upon arrival, participants were asked to change into their athletic wear and various measures were taken (i.e. weight, age, leg length, knee width, etc.). Standing with both feet on either side-by-side force plates, at comfortable and natural stance width, the participants were asked to self-initiate gait (i.e. no start cue was given). Practice was allowed in order to ensure targeting did not take place. First, participants initiated gait with their right limb, stepping on the third force plate with their first step, and continuing to the end of the walkway. Then, subjects were asked to initiate gait with their left limb in the same manner as with their right limb. Five trials with each limb leading were collected. Subjects were informed that they could rest at any time and all improper trials were deleted and collected once again.

2.4.Outcome parameters

Data collected from the three force platforms was exported and the CoP_{net} was calculated across all three force platforms. Parameters of CoP_{net} A/P displacement in cm (A1, A2, A3), CoP_{net} M/L displacement in cm (M1, M2, M3) and duration of phase in seconds (T1, T2, T3) for each APA phase were calculated. As well, the total APA CoP_{net} A/P displacement in cm, CoP_{net} M/L displacement in cm and duration in seconds (A_{total}, M_{total}, T_{total}) were calculated as the difference from quiet standing to the end of APA3 phase and tabulated. The APA4 phase describes the end of the APA, as the trailing limb leaves the ground, and the CoP_{net} pattern of APA4 is omitted from the current analysis. The current APA calculations are modelled according to the cogent research carried out by Cau and colleagues [30]. The main outcome paramters were established as A_{total} and T_{total}.

2.5.Statistical analysis

All statistical analyses were carried out using SPSS 24 (IBM Corp., NY). Level of significance was set at $p \leq 0.05$. A non-parametric analysis of variance was carried out via a Friedman related samples test with a Bonferroni correction. A pairwise comparison was done in order to control for age. Further, Wilcoxon related sample tests were carried out between conditions for each parameter (i.e. control, intact limb, prosthetic limb). Statistical design compared both conditions (i.e. gait initiation with the right and left leading limb) in control subjects. For instances when the right and left leading limb conditions were not statistically different in controls, the average result of the right and left limbs were pooled together to form one control value. In DTTA subjects, gait initiation with the prosthetic versus intact limb was compared and the results obtained in the control group were compared to those obtained in the DTTA group (i.e. control, intact limb, prosthetic limb). Finally, effect size was also calculated between conditions (i.e. control, intact, prosthetic) using Cohen's d ($d \geq 0.5$ = moderate clinical significance and $d \geq 0.8$ = strong clinical significance).

3. Results

A typical APA CoP_{net} displacement pattern for all conditions (i.e. control, prosthetic and intact limb leading) is displayed in Figure 2. The results with regards to the mean and standard deviation of the various APA phase parameters are outlined in Table 2 and Figure 3. It was thought most appropriate to display the mean and standard deviation values for A/P and M/L CoP_{net} displacements for each APA phase as such in order for comparison to be made with regards to direction. Because results were not statistically different between the right and left limb in control subjects, the mean results for the right and left control limbs was used for analysis. As well, conditions will be discussed with regards to the trailing limb.

--------------------------------Insert Figure 2 approximately here--------------------------

---------------------------------Insert Table 2 approximately here----------------------------

---------------------------------Insert Figure 3 approximately here---------------------------

When considering the A/P CoP_{net} displacement, no significant differences were observed for A1 and A3 between all three conditions (i.e. controls, intact and prosthetic limbs). A significant increase in anterior CoP_{net} displacement was observed in the prosthetic limb when compared to the control ($p=0.01$) and intact limb ($p=0.04$) was observed for A2, but no significant difference was seen between controls and the intact limb. For A_{total}, a significant difference was observed between the control and prosthetic limb ($p=0.05$) as an anterior total CoP_{net} displacement was observed in the prosthetic limb when compared to a posterior total CoP_{net} displacement observed in controls. A reduced posterior total CoP_{net} displacement was observed in the intact limb but this was not significantly different when compared to the control or prosthetic limbs.

No significant differences were found for the M/L CoP_{net} across all three conditions for M1, M2 and M_{total}. A significant reduction in M/L CoP_{net} displacement was observed in the trailing intact limb when compared to both the control ($p=0.05$) and the prosthetic ($p=0.04$) limb in M3. Finally, no significant difference was observed between the control and prosthetic limb for M3.

The time taken to complete the total APA and the various phases of APA showed no significant differences for T1 and T3 between all three conditions. A significantly reduced time was observed in controls when compared to the intact limb ($p=0.01$) and the prosthetic limb ($p=0.01$) for T2, but no significant difference was observed between the intact and prosthetic limb for T2. For the T_{total} a significantly reduced time was observed in the control limb when compared to the prosthetic limb ($p=0.05$). For all time APA parameters, no significant differences were observed between the intact and control or prosthetic limbs.

4. Discussion

The DTTA and TTA present distinct SSWV and quiet standing profiles, the DTTA having further reductions in stability and propulsion when compared to the TTA. However, in gait initiation, no studies have yet explored the APA in the unique population of the DTTA though gait initiation is a complex task which requires synchronous activation of several control systems. The aim of the current study was to compare the APA pattern (i.e. CoP_{net} displacement and time of APA) employed by DTTA to age-matched controls. The APA strategy was divided between 12 parameters across total and three phases of APA in three groups (i.e. control, intact limb, amputated limb). Significantly increased T_{total} was observed in the prosthetic trailing limb when compared to controls (1.11±0.18s and 0.81±0.18s, respectively; $p=0.05$), significantly reduced M3 CoP_{net} displacement was observed in the intact trailing limb when compared to both the control and prosthetic trailing limb conditions (3.59±0.29cm vs. 7.03±2.00cm and 7.11±0.40cm, respectively; $p=0.05$ and $p=0.04$) and a significant difference in anterior A_{total} CoP_{net} displacement was observed in the prosthetic trailing limb condition when compared to the posterior A/P_{total} CoP_{net} displacement observed in the control trailing limb conditions (anterior 3.20±2.96cm vs. posterior 3.57±1.93cm, respectively; $p=0.05$). Limitations of this study lie in the number of participants and heterogeneity of participants mainly with regards to age, but also physical fitness levels and perhaps concurrent dysvascularity which were not measured. Results of this study seem to indicate that the anterior A_{total} CoP_{net} displacement observed beneath the prosthetic trailing limb would be specific to the DTTA population as other studies carried out in TTA, all amputation causes confounded, found both anterior and reduced posterior A_{total} CoP_{net} displacement. The implications of these findings would support an even greater need to differentiate between the distinct populations of DTTA and TTA, as well as support the idea of balance over propulsion in the specific DTTA population.

4.1 A/P CoP_{net} displacement

A significant difference in A_{total} CoP_{net} displacement was observed between an anterior 3.2±2.96cm in the prosthetic trailing limb condition when compared to a posterior 3.57±1.92cm observed in controls (p=0.05). No significant differences were observed between the posterior A_{total} CoP_{net} displacement in the intact trailing limb condition (2.15±3.00cm) and the control or prosthetic trailing limb conditions (p=0.24 and p=0.16, respectively).

For efficient propulsion during gait initiation, a sufficient posterior CoP_{net} displacement on the trailing limb is required as this produces a torque propelling the total CoM forward towards the first step. Posterior CoP_{net} displacement during gait initiation in control subjects have been shown to produce a sufficiently large torque effect to propulse the total CoM forward, posterior shift having found to be approximately 3.5 to 4.7 cm and 3.2 to 3.5 cm in healthy and older adults, respectively [2]. The current findings in our control group are therefore in line with these results.

The reduced posterior CoP_{net} displacement in the intact loaded limb observed in the current study (posterior 2.15±3.00cm), though not significantly different when compared to controls (posterior 3.57±1.92cm; $p=0.24$), has also been found in prior studies where all reasons for amputation in the TTA were included [8,15,16,17,26]. However, when the prosthetic limb was loaded (i.e. trailing limb), two studies corroborate an anterior CoP_{net} displacement [16,17]. Other studies have observed a small and reduced posterior CoP_{net} displacement when compared to able-bodied and trailing intact limb [8,15,26].

The anterior total CoP_{net} shift observed beneath the trailing prosthetic limb would appear to be an important result within the population of DTTA. As stated, previous studies have found both anterior and reduced posterior total CoP_{net} shift [8,15,16,17,18]. Because prior studies investigated APA in both traumatic and dysvascular TTA, the mixed results obtained are perhaps due to this. Indeed, the current study observed an anterior displacement (anterior 3.2±2.96cm vs. posterior 3.57±1.92cm in controls; $p=0.05$) in all but 2 participants. Though no measurements were taken with regards to physical fitness levels, the two participants who displayed a reduced posterior shift (rather than anterior) were two younger individuals with an active lifestyle, arguably the most physically fit DTTA participants. When traumatic and dysvascular TTA are compared, DTTA are deconditioned and less active [21,22]. The total anterior CoP_{net} shift observed would appear to be related to further reductions in stability caused by reduced physical fitness levels, sensory loss, peripheral dysvascularity in the intact limb, etc. associated with DTTA.

By producing a total anterior CoP_{net} shift, the DTTA are able to maintain a more stable position as the CoP_{net} lies closer to the CoM. That is, the closer the CoM lies in relation to the CoP_{net}, the more stable the position. Indeed, as CoP_{net} shifts posteriorly, the CoM is pushed further forward, creating an unstable situation which propels the body forward for gait initiation [2]. By producing an anterior CoP_{net} shift in DTTA, the CoM is not pushed as far forward, maintaining a more stable and secure situation but an inefficient propulsive thrust. Because the first step, with the leading intact limb, has been initiated by toe-off, the CoM has progressed forward to a position about midway from the first step (i.e. needing to be caught by the leading limb to avoid falling), gait initiation is possible, though propulsion is reduced by the APA strategy beneath the trailing intact limb, and moreover, in the trailing prosthetic limb.

4.2 M/L CoP_{net} displacement

The results of the current study showed a significantly reduced M3 CoP_{net} displacement in the intact trailing limb condition (3.59±2.93cm) when compared to the control and prosthetic limb conditions (7.03±2.00cm and 7.11±4.01cm, respectively; $p=0.05$ and $p=0.04$, respectively). No significant differences were found between control, intact and prosthetic trailing limb conditions for M1 (6.42±3.82cm, 6.78±4.67cm and 3.91±1.83cm, respectively; all $p>0.05$), M2 (11.29±5.9cm, 16.21±9.89cm and 13.60±5.39cm, respectively; all $p>0.05$) and M_{total} (18.62±17.65, 19.53±9.60 and 20.23±2.84cm, respectively; all $p>0.05$).

The limb load asymmetry observed in quiet standing between the intact and prosthetic limb has been well documented. The traumatic and DTTA place significantly more weight on the intact limb when compared to the prosthetic limb due to reduced strength and stability in the prosthetic limb [21,25,27]. This limb load asymmetry was observed solely in the M3 phase of the M/L CoP_{net} displacement of the trailing intact limb condition. As the weight transfers from the prosthetic leading limb to the trailing intact limb, and the prosthetic limb reach toe-off into the first step, the weight transfer to the trailing intact limb travels from a position closer to the trailing intact limb, causing the reduced M/L CoP_{net} displacement observed when compared to the trailing prosthetic and control conditions.

Surprisingly, no significant differences were observed for M1, M2 and M_{total} across all conditions (i.e. control, intact and prosthetic trailing limb conditions). As discussed, load limb asymmetry causes the quiet standing CoP_{net} to be placed towards the intact limb rather than midway. This is perhaps due to the large standard deviations observed in the data.

4.3 Time of CoP_{net} displacement

The results of the current study showed a significant difference for T2 for control trailing limb condition (0.19±0.03s) when compared to intact and prosthetic trailing limb conditions (0.32±0.06s and 0.39±0.11s, respectively) (both $p=0.01$). As well, for T_{total} a significant difference was observed between the control and prosthetic trailing limb condition (0.81±0.18s and 1.11±0.18s, respectively; $p=0.05$). No significant differences were observed between the intact and prosthetic trailing limb conditions for T2 ($p=0.11$) and for the intact trailing limb condition when compared to both the control and prosthetic trailing limbs for T_{total} ($p=0.07$ and $p=0.33$, respectively). Finally, no significant differences were observed between control, intact and prosthetic trailing limb conditions in T1 (0.49±0.16s, 0.56±0.25s and 0.58±0.19s, respectively; all $p>0.05$) and T3 (0.13±0.04s, 0.12±0.06 and 0.13±0.05, respectively; all $p>0.05$)

The increased total time of CoP_{net} displacement observed in the prosthetic limb, when compared to controls, has been proposed as a 'movement time' strategy in TTA in order to counteract the stability and propulsion limitation imposed by the prosthetic limb. The results of the current study with regards to DTTA support this theory in this specific population of DTTA [16]. Indeed, as sensory information and structure (i.e. bone and muscle) are missing in the prosthetic limb, stability on then prosthetic limb is reduced and precarious. It is well documented that the TTA prefer to initiate gait with the prosthetic limb, as the total body weight is then loaded on the intact limb [17]. In the condition where the intact limb is leading, total body weight must be loaded to the prosthetic limb in order for the intact limb to be unloaded and swing into the first step. This has been shown to take more time in the TTA compared to controls [25]. In doing so, an unsteady and precarious situation is created for the TTA. In order to ensure stability, and reduce risk of falling, TTA choose to favour stability over propulsion, thus taking more time to initiate gait [17]. This favouring of stability versus propulsion is also highlighted by the increase in total time taken for the APA when initiating gait with the intact limb (i.e. trailing prosthetic limb) when compared to controls.

As discussed with regards to M/L CoP_{net} displacement, the limb load asymmetry in quiet standing, as additional weight bearing is placed on the intact limb, is again reason for the significantly increased time observed in the DTTA when compared to the control (0.19±0.03s for controls vs. 0.32±0.06s in intact and 0.39±0.11s in prosthetic limb; both $p=0.01$ when compared to controls) in T2 and between controls and the prosthetic limb leading for T_{total} (0.81±0.18s and 1.11±0.18s, respectively; $p=0.05$). More time is needed to unweight the additional body weight put on the intact limb to the prosthetic limb.

4.4 APA in the DTTA

The most important alteration to the gait pattern seen in TTA is due to the missing ankle joint and associated musculature [28]. Indeed, this missing joint has implications not only in gait, but in gait initiation as well. As stated above, the plantarflexor action, created by the gastrocnemius-soleus complex, is the most important power generator in walking. The absence of this muscle complex in TTA prosthetic limb leads to important power reductions and important changes to gait initiation are therefore observed.

The APA strategies with regards to reduced time and the anterior CoP_{net} observed in the current study make evidence of a careful gait initiation pattern in the DTTA. Prior studies have discussed this strategy employed by TTA [17,29]. Because gait initiation poses important challenges to balance, the TTA choose to control for this precarious situation by adopting a careful strategy, favouring balance over propulsion. The results with regards to the reduced and anterior CoP_{net} displacement and increased time taken for the APA observed in the current study support this notion. As well, the TTA, and DTTA in this current study, favour the more stable gait initiation technique by leading with their prosthetic limb, evidence again of a careful gait initiation strategy [16,17].

4.5 Limits, implications, clinical significance & future work

Limitations of this study lie mainly in the number of participants and heterogeneity of participants mainly with regards to age but also physical fitness levels and perhaps concurrent dysvascularity which were not measured. The results obtained make proof of possible outliers as the standard deviations are quite large: statistical power for many parameters is lacking conceivably due to large standard deviation values. Though the number of participants in the current study reflect those of prior and similar studies, the lack of heterogeneity within the TTA and DTTA population is a challenge in itself [16,17]. Heterogeneity of participants is seen mainly in other dysvascularity issues (i.e. in intact limb, eyesight, residual limb, etc.), age as well as overall physical fitness levels. The participants of the current study varied largely with regards to age, as well.

Another limitation to this study lies in the choice to average the right and left limb values obtained in the control subjects. Though no statistical differences were found between the right and left limbs, and though the mean and standard deviation values were observed to ensure no outliers, it is possible that future studies comparing right and left limbs distinctly in controls when compared to TTA would allow not only to observe differences in the TTA, but also be able to report on dominant limb and the role of the dominant limb in gait initiation.

For those results statistically different, a large clinical significance was found for T2 between control and intact limbs (d =4.33) and between the control and prosthetic limbs (d =6.67), for T_{total} between the control and prosthetic limbs (d =1.67), for A2 between control and prosthetic limbs (d =4.00) and between intact and prosthetic limbs (d =1.96), for A_{total} between the control and prosthetic limbs (d =3.51) and finally, for M3 between the control and intact limbs (d=1.72) and between the intact and prosthetic limbs (d=1.21). These large values obtained are far above the 0.8 limit established as the threshold for large clinical significance. Indeed, that these variables are clinically significant adds to the significantly different results obtained between the above conditions and bear witness to the very important difference in APA strategy in the DTTA.

Further, some parameters demonstrated large clinical significance, yet were not statistically different: for T2 between the intact and prosthetic limbs (d=1.17), for T_{total} between the control and intact limbs (d =1.06), for A1 between the control and intact limbs (d =0.84) and between the control and prosthetic limbs (d = 0.80), for A3 between the control and intact limbs (d = 0.73), for A_{total} between the control and intact limbs (d =0.73) and between the intact and prosthetic limbs (d =1.81) and finally, for M2 between the control and intact limbs (d =0.83). Thus, future work is needed to further understand the differences between these variables in the DTTA when compared to healthy adults. Though these parameters were clinically significant, the importance of these differences cannot fully yet be understood: further studies are needed to fully investigate and understand these clinically significant differences. As mentioned, it is thought that the number of participants as well as the heterogeneity of participants is responsible for large standard deviations and thus, further investigation is needed.

Further work is needed to understand the differences in gait initiation between the dysvascular and traumatic TTA. As well, future studies are warranted to understand the gait termination strategies utilized by the specific DTTA population. Research into gait initiation must then be disseminated to the rehabilitation and care setting. Because instability is particularly high during such tasks as gait initiation, transitioning from bi- to mono-pedal stance, improvements with regards to intact limb gait initiation is needed to improve response to everyday perturbations.

5. Conclusion

To produce gait initiation, the DTTA utilize different APA strategies when compared to controls. Through increased APA time and anterior or reduced CoP_{net} displacement beneath the trailing prosthetic limb reduce propulsive torque for gait initiation, it allows for a more careful gait initiation, favouring stability over propulsion, in the DTTA as prior observed in the TTA. Perhaps the most important result from the current study is the anterior CoP_{net} displacement observed beneath the trailing prosthetic limb within the specific DTTA population as it bears witness to the further reductions in ambulation capacity brought upon by dysvascular amputation. Rehabilitation should focus on improving gait initiation with the intact limb in order to prepare for unexpected everyday perturbations.

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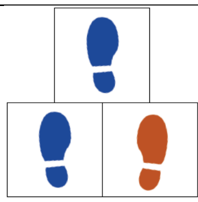
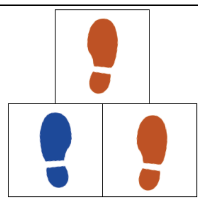
Tables

Table 1- Mean ($\bar{x}$) ± standard deviation & median (M) (minimum: maximum) participant characteristics for both experimental groups.

	Control group	TTA group
N	10	10
Age	$\bar{x}$ = 57.7 ± 16.4 yrs old M= 61 (25: 81)	$\bar{x}$ =59.1 ± 17.3 yrs old M= 61.5 (25: 88)
Sex	8M: 2F	7M: 3F
BMI	$\bar{x}$ = 25.7 ±4.2 kg/m^2 M= 24.6 (21.5: 33.3)	$\bar{x}$ =27.0 ±7.5 kg/m^2 M= 27.0 (19.9: 35.4)
Years since amputation		$\bar{x}$ = 3.8 ± 7.8 yrs M= 1 (1: 26)

Table 2- Mean ($\bar{x}$) ± standard deviation & median (M) (minimum: maximum) of time, A/P and M/L CoP_{net} displacement values for APA and total APA phases beneath the identified trailing limb for controls and DTTA. Blue footprints outline the intact limb and red footprints outline the prosthetic limb in the diagrams below. (* † denote significant differences between conditions for a single parameter; negative values denote an anterior direction displacement)

	CONTROL	 INTACT trailing limb	 PROSTHETIC trailing limb
T1 (s)	$\bar{x}$ = 0.49 ± 0.16 M= 0.54 (0.27: 0.70)	$\bar{x}$ = 0.56 ± 0.25 M= 0.56 (0.22: 1.08)	$\bar{x}$ = 0.58 ± 0.19 M= 0.56 (0.39: 0.90)
T2 (s)	$\bar{x}$ = 0.19 ± 0.03* † M= 0.20 (0.14: 0.24)	$\bar{x}$ = 0.32 ± 0.06* M= 0.33 (0.23: 0.42)	$\bar{x}$ = 0.39 ± 0.11 † M= 0.40 (0.25: 0.58)
T3 (s)	$\bar{x}$ = 0.13 ± 0.04 M= 0.12 (0.09: 0.21)	$\bar{x}$ = 0.12 ± 0.06 M= 0.14 (0.04: 0.19)	$\bar{x}$ = 0.13 ± 0.05 M= 0.14 (0.06: 0.18)
T_{total} (s)	$\bar{x}$ = 0.81 ± 0.18* M= 0.85 (0.55: 1.08)	$\bar{x}$ = 1.00 ± 0.26 M= 0.95 (0.69: 1.53)	$\bar{x}$ = 1.11 ± 0.18* M= 1.12 (0.89: 1.42)
A1 (cm)	$\bar{x}$ = 3.47 ± 1.64 M= 2.99 (1.61: 5.49)	$\bar{x}$ = 2.09 ± 1.63 M= 1.43 (0.85: 5.77)	$\bar{x}$ = 2.16 ± 1.00 M= 2.27 (0.75: 3.72)
A2 (cm)	$\bar{x}$ = 2.31 ± 0.83* M= 2.10 (1.19: 3.81)	$\bar{x}$ = 2.12 ± 1.72 † M= 1.41 (2.99: 4.99)	$\bar{x}$ = 5.51 ± 2.06 *† M= 5.34 (2.48: 8.43)
A3 (cm)	$\bar{x}$ = 2.49 ± 0.94 M= 2.51 (1.11: 3.66)	$\bar{x}$ = 1.79 ± 1.95 M= 1.01 (0.12: 5.51)	$\bar{x}$ = 2.52 ± 2.12 M= 1.87 (0.62: 5.96)
A_{total} (cm)	$\bar{x}$ = 3.59 ± 1.92 * M= 3.67 (1.35: 6.10)	$\bar{x}$ = 2.15 ± 2.96 M= 2.15 (-0.03: 5.90)	$\bar{x}$ = -3.19 ± 2.96 * M= -2.46 (-4.44: 0.42)
M1 (cm)	$\bar{x}$ = 6.42 ± 3.82 M= 6.34 (1.40: 14.31)	$\bar{x}$ = 6.78 ± 4.67 M= 5.40 (1.39: 15.23)	$\bar{x}$ = 3.91 ± 1.83 M= 3.44 (1.56: 7.11)
M2 (cm)	$\bar{x}$ = 11.29 ± 5.91 M= 8.60 (6.76: 24.66)	$\bar{x}$ = 16.21 ± 9.89 M= 15.33 (3.91: 37.65)	$\bar{x}$ = 13.60 ± 5.39 M= 13.34 (6.21: 23.46)
M3 (cm)	$\bar{x}$ = 7.03 ± 1.99 * M= 6.96 (4.07 : 9.52)	$\bar{x}$ = 3.59 ± 2.93 *† M= 3.76 (0.24: 8.17)	$\bar{x}$ = 7.11 ± 4.01 † M= 7.99 (0.36: 13.31)
M_{total} (cm)	$\bar{x}$ = 18.62 ± 7.80 M= 16.55 (10.83: 36.02)	$\bar{x}$ = 19.53 ± 9.60 M= 18.78 (7.44: 37.76)	$\bar{x}$ = 20.22 ± 2.84 M= 20.31 (15.27: 23.46)

Figures

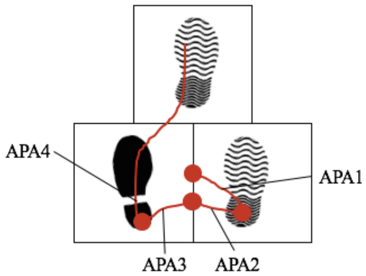


Figure 1- Sketch of CoP_{net} displacement across 3 force platforms of experimental set-up. The APA phases are outlined: APA1, APA2, APA3 & APA4. Note: left foot is trailing limb in this figure.

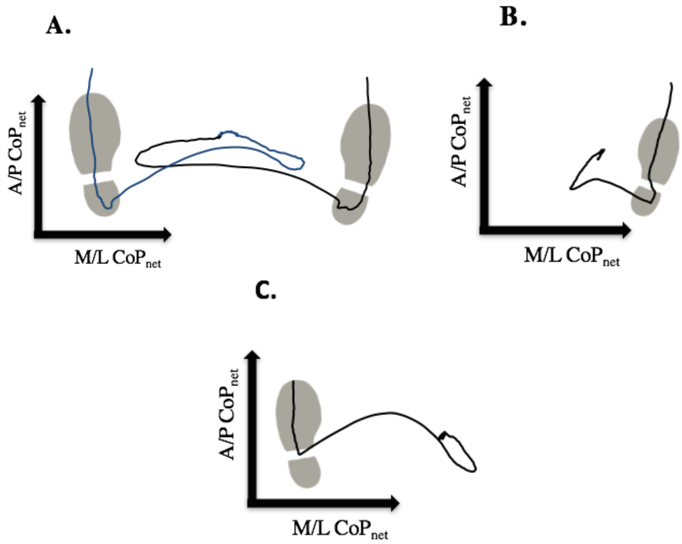


Figure 2- Example of typical CoP_{total} displacement in A. Controls; B. Intact limb trailing; C. Prosthetic limb trailing. The outlined foot represents the trailing limb.

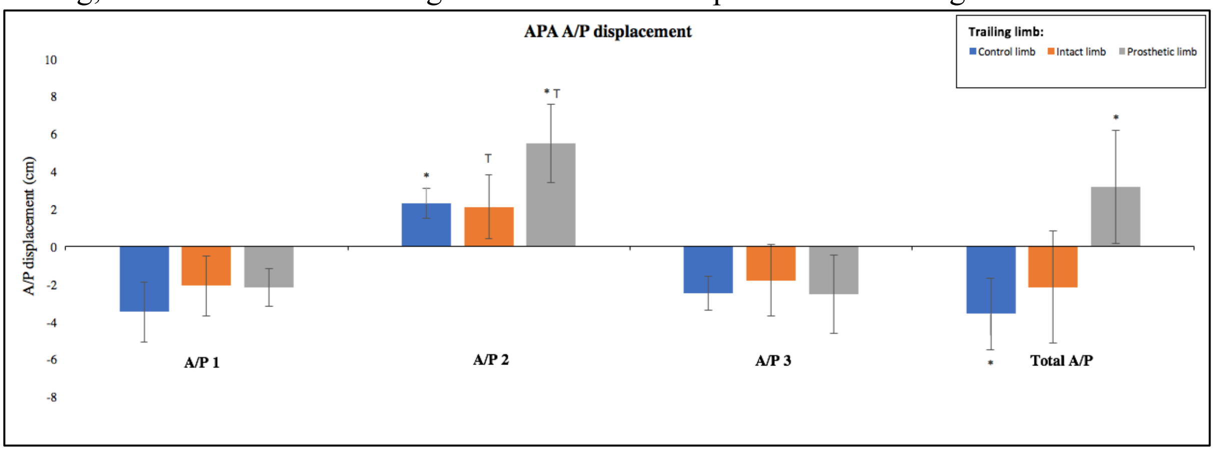


Figure 3- Mean (± standard deviation) in A/P direction. APA CoPnet displacements across all three phases of APA and total displacement in controls, intact and prosthetic limbs. Positive values describe an anterior A/P displacement.

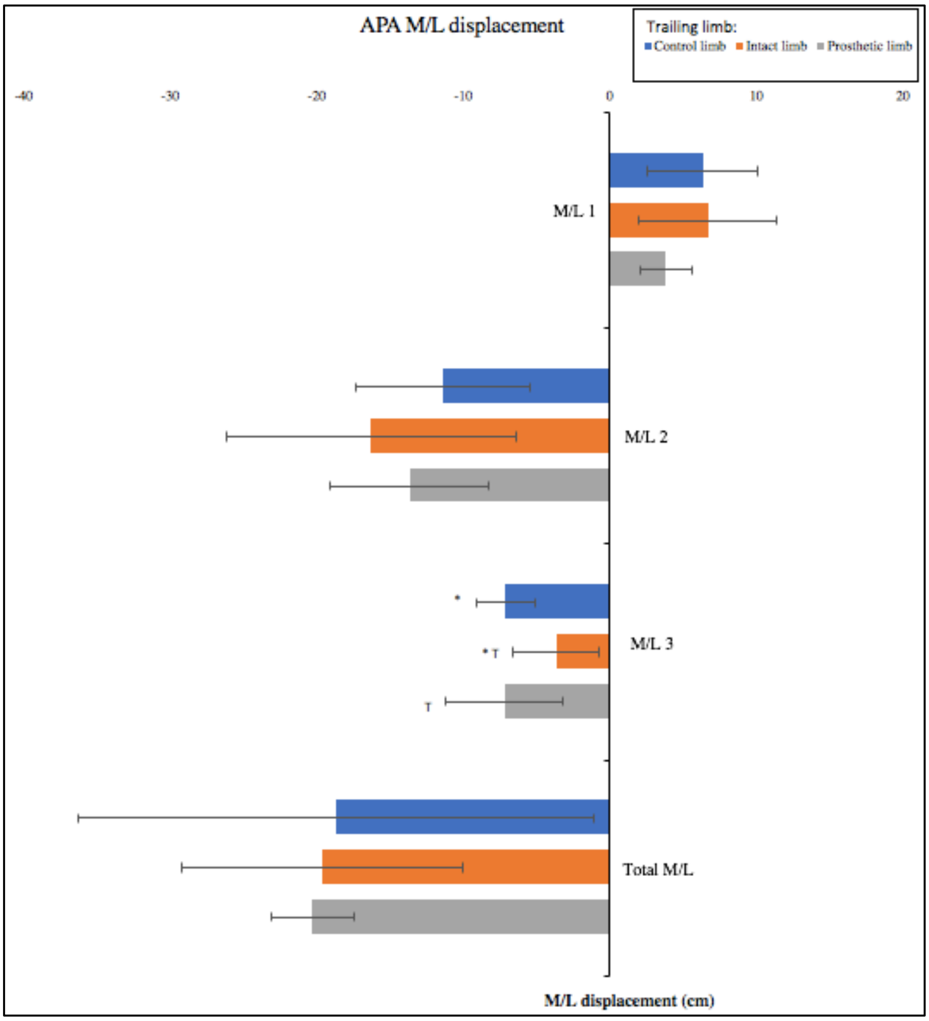


Figure 4- Mean (± standard deviation) in M/L direction. APA CoPnet displacements across all three phases of APA and total displacement in controls, intact and prosthetic limbs. Positive values describe a M/L displacement towards the leading limb.

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Chapter 5: Article III

Underlying gait initiation mechanisms in the dysvascular transtibial amputee.

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Abstract

The dysvascular transtibial amputee is the most sizeable and growing population in the United States. Though gait initiation is well understood in the healthy adult, few studies have been carried out in the transtibial amputee. Moreover, no studies have focused solely on gait initiation in the dysvascular transtibial amputee population. Gait initiation precedes every walking bout, is part of almost all activities of daily living and poses important constraints to balance as individuals must shift from bi-pedal to mono-pedal stance, from static to forward motion. Thus, the purpose of the present study was to compare the underlying biomechanical differences in the gait initiation parameters of dysvascular transtibial amputees with those of healthy age-matched controls. Ten dysvascular transtibial amputees and ten controls participated in this study consisting of five gait initiation trials with the right limb, from quiet standing to steady-state walking velocity, followed by five gait initiation trials with the left limb. Kinetic and kinematic data was recorded for seven parameters. A reduced steady-state walking velocity was observed in the dysvascular transtibial amputee (1.07±0.2 m/s vs. 1.30±0.2 m/s in control subjects; p=0.03), as expected due to the increased braking force (-0.3±0.2 N/kg vs. -0.5±0.3 N/kg; p=0.03) and reduced propulsive impulse (2.0±0.9 N•s/kg vs. 3.9±0.6 N•s/kg; p=0.03) observed in the prosthetic limb when compared to controls. The propulsive impulse possible by the prosthetic limb makes evidence of gluteal contribution to gait initiation propulsion in the first step, in the absence of the gastrocnemius-soleus muscle complex. Additionally, no difference was observed between intact and control limb vertical force (91.2±7.1 vs. 96.8±4.1 N/kg, respectively; p=0.18). This may be a protective mechanism in the dysvascular transtibial amputee, leading to reduced osteoarthritis risk in the intact limb. These results corroborate the notion of '*careful*' gait initiation in the dysvascular transtibial amputee, as stability is favored over propulsion, contributing to a reduced steady-state walking velocity. The implications of these findings

make proof that gait initiation leading with both limbs should be an avid focus in dysvascular transtibial amputee rehabilitation, as this complex motor task is a critical component of daily living and function.

1. Introduction

In the United States, it is estimated that over 1.6 million people are living with an amputation and of this, over half are amputated in the lower limb for dysvascular reasons[1]. As well, trends indicate that the number of amputations is projected to increase, in large part because of the increasing number of individuals affected by dysvascular diseases, namely Diabetes[2]. Finally, statistics indicate that the most common amputation is the transtibial amputation[3].

Following amputation, transtibial amputees (TTA) must relearn walking and many confounding factors interplay to impact the gait pattern. Indeed, walking necessitates an increased demand in energy and thus quality of life is significantly diminished [4,5].

Because walking is the most common form of locomotion and it is part of almost all activities of daily living[6,7], the ability to walk is an indicator of overall health and it dictates autonomy[8]. Early studies have looked at the walking pattern in the TTA population and authors have concluded that the walking pattern is altered. Indeed, TTA walk at a significantly reduced walking velocity when compared to able-bodied individuals [5,9]. Both kinetic and kinematic patterns are altered, principally due to the missing foot and ankle joint, the gastrocnemius and soleus muscle complex being the major propulsors when walking in the healthy individual [10,11,12,13].

Gait initiation precedes every walking bout and relies on an intricate interplay between various systems to be achieved. Gait initiation is a complex motor task, passing from quiet standing static state to a dynamic state, transferring weight from bi-pedal in quiet standing to mono-pedal in initiating the first step of walking. To reach steady-state walking velocity (SSWV), acceleration must be provided to the body center of mass (CoM). The propulsive force must be greater than that of braking force to produce this acceleration. Therefore, the propulsion produced by the CoM falling forward as well as the soleus and gastrocnemius muscle contraction at push-off are important in producing the SSWV [14,15,16,17].

Studies have shown that in gait initiation, the TTA prefer to lead with their prosthetic limb and load the trailing intact limb to initiate gait, a more stable situation. As well, the TTA take more time to initiate gait [18,19,20] and an important reduction in gait initiation velocity was observed in TTA when compared to controls [21,22].

Though gait initiation has been amply explored in the healthy adult, few studies have been undertaken in the TTA population and no studies have been carried out solely in the dysvascular TTA (DTTA). It is known that the DTTA poses important constraints to biomechanics when compared to the traumatic TTA as often confounding health factors interplay with amputation (i.e. presence of peripheral dysvascularity in the non-amputated “intact” limb, sensory loss, vision loss, physical deconditioning, etc.) [23,24]. Indeed, survival rate is lower in the DTTA population when compared to traumatic TTA[25]. When considering SSWV, there is an overall reduced SSWV in DTTA when compared to traumatic TTA, the VO_{2max} demand being increased [26]. Studies have investigated postural control during quiet standing in the DTTA compared to the traumatic TTA. Increased sway and inability to balance in single limb stance on the prosthetic limb were the significant differences observed in the DTTA when compared to the TTA [23,27].

Thus, the purpose of the present study is to compare the underlying kinetics of the first step in the gait initiation parameters of DTTA with those of healthy age-matched controls. To our knowledge, this is the first gait initiation study to solely include DTTA.

2. Methods

Both the methodology and participant characteristics of the current study have been presented in a prior study, Roberts & Prince (2018) describing the anticipatory postural adjustments prior to and during gait initiation. The current study analyzes the data of the first step kinetics during gait initiation. Thus, to ensure proper comprehension to the reader in light of this new topic, the methods and participants are presented.

2.1 Subjects

A total of 10 subjects with a unilateral DTTA were recruited via the Institut de réadaptation Gingras-Lindsay de Montréal. A group of 10 control subjects were recruited via acquaintances of the researchers at l'Université de Montréal. The control subjects were healthy adults, age-matched to the DTTA subjects.

Subject characteristics are summarized in Table 1 found in a prior work by Roberts & Prince (2018). Any other conditions and/ or diseases which could have an impact on the standing and locomotor pattern (i.e. other than that having caused amputation, for example Type II Diabetes) were reason for subject exclusion. All DTTA subjects wore their own prosthetic device for testing and all prostheses were equipped with a passive foot. All subjects provided informed consent prior to testing. This study was approved by the Comité d'éthique de la recherche en santé de l'Université de Montréal and the Comité d'éthique en recherche des établissements du Centre de recherche interdisciplinaire de réadaptation du Montréal métropolitain.

2.2 Equipment

A walkway, with three embedded AccuGait force plates (Advanced Medical Technology Inc., MA), was set up surrounded by 8 Flex13 motion capture cameras from the OptiTrack motion analysis system (NaturalPoint Inc., OR). A total of 39 reflective markers were placed on the subject at anatomical landmarks based on the Plug-in Gait model (Vicon Motion Systems Ltd., UK).

Subjects were asked to walk looking straight ahead during each trial to avoid targeting of the force plates and practice was allowed in order to ensure targeting did not take place. The force plates measured the ground reaction forces (GRF) in all three planes of movement (vertical, anteroposterior (A/P) and mediolateral (M/L). Both kinetic and kinematic systems were synchronized and sampled at 100Hz. All data analysis was carried out using a MATLAB program (The MathWorks Inc., MA) created for the purpose of the present research project.

2.3 Procedure

Upon arrival, subjects changed into their athletic attire, various subject measures were taken (i.e. weight, age, leg length, knee width, etc.) and reflective markers were then fastened to skin, prosthesis and/or tight-fitted clothing. Subjects were first asked to self-initiate gait (i.e. no start cue was given) with their right limb, from quiet standing with each foot on force plates 1 and 2 (Figure 1), naturally stepping onto the third force plate with their first step and continuing to the end of the walkway. Subjects were then asked to initiate gait with their left limb in the same manner. Five trials with each limb leading were collected. Subjects were informed that they could rest at any time and all improper trials were deleted and collected once again.

--------------------------------Insert Figure 1 approximately here---------------------------

2.4 Outcome parameters

From the data collected, the heel contact and toe-off events were identified and gait velocity was calculated. The maximum braking, propulsive and vertical forces as well as associated impulses and loading rate of the first step were computed. Impulses were calculated as the time-force integral and loading rate as the slope of the force (i.e. force divided by time) leading to the maximum vertical force. Force and impulse parameters were divided by body weight and time normalized to 100% of stance phase. The main outcome parameters were established as SSWV and propulsive impulse.

2.5 Statistical analysis

All statistical analyses were carried out using SPSS 24 (IBM Corp., NY). Level of significance was set at $p < 0.05$. A non-parametric analysis of variance was carried out via a Friedman related samples test with a Bonferroni correction. A pairwise comparison was done in order to control for age. Further, Wilcoxon related sample tests were carried out between conditions for each parameter (i.e. control, intact limb, prosthetic limb). Statistical design compared both conditions (i.e. gait initiation with the right or left leading limb) in control subjects. For instances when the right and left leading limb conditions were not statistically different in controls, the average result of the right and left limb were pooled together to form one average control value. In DTTA subjects, gait initiation with the prosthetic versus intact limb was compared. Then, the results obtained in the control group were compared to those obtained in the DTTA group (i.e. control, intact limb, prosthetic limb). Finally, effect size was also calculated between conditions (i.e. control, intact, prosthetic) using Cohen's d ($d \geq 0.5$ = moderate clinical significance and $d \geq 0.8$ = strong clinical significance).

3. Results

Because all results were not statistically different between the right and left limbs in the control group, the mean of the right and left limb results were taken and combined to one control value for all seven parameters observed. The average SSWV (1.30 ± 0.2 m/s) attained by the control group was achieved on average at step 3 with both right and left leading limbs. Average SSWV achieved by the DTTA group was significantly reduced (1.07 ± 0.2 m/s) when compared to controls and on average was reached at step 4. No significant difference existed between the SSWV with either the intact or prosthetic leading limb. As well, there was no significant difference between the number of steps taken to reach SSWV in controls and DTTA.

The average vertical and A/P GRF profiles of the first step were plotted and the various force and impulse parameters calculated are displayed in Table 1. Maximum braking and propulsive forces and impulses as well as vertical force and loading rate were also computed and are displayed in Table 1.

The maximum braking force was found to be significantly greater in the intact limb when compared to the prosthetic limb ($p=0.03$), but no significant differences existed between the prosthetic and control limb ($p=0.5$) or the control and intact limb ($p=0.4$). No significant differences were observed between all three conditions (control, intact and prosthetic limb leading) for the maximum propulsive force.

The propulsive impulse in the prosthetic limb was found to be significantly reduced when compared to the intact ($p=0.03$) and control limb ($p=0.03$). No significant difference in propulsive impulse existed between the intact and control limb ($p=0.2$). No significant differences were observed either between all three conditions for the braking impulse (all $p>0.05$).

The maximum vertical force in the prosthetic limb was found to be significantly reduced when compared to the intact ($p=0.04$) and control limb ($p=0.03$). No significant difference in maximum vertical force was observed between the intact and control limb ($p=0.2$). The loading rate was found to be significantly reduced in the prosthetic limb when compared to both the intact ($p=0.02$) and control limb ($p=0.01$) while no significant difference was observed for the loading rate between the intact and control limb ($p=0.4$).

---------------------------------Insert Table 1 approximately here-----------------------------

4. Discussion

The first step allows for 75-90% of the total SSWV to be achieved in the healthy adult. As well, the DTTA and TTA present different quiet standing and SSWV profiles, yet distinction in gait initiation has yet been studied. Thus, the purpose of the current study was to compare the underlying biomechanical differences in the gait initiation parameters between DTTA and healthy age-matched controls. Seven gait initiation parameters were observed in three gait initiation conditions (i.e. leading control, intact and prosthetic limb). The results confirmed a reduced SSWV in the DTTA when compared to controls (1.07±0.2m/s and 1.30±0.2m/s, respectively; p=0.03) as shown in previous studies. Maximum braking force was significantly reduced in the prosthetic limb when compared to the intact limb (-0.3±0.2N/kg and -0.6±0.2N/kg, respectively; p=0.03) and propulsive impulse was significantly reduced in the prosthetic limb when compared to both the control and intact limbs (2.0±0.9N•s/kg vs. 3.9±0.69N•s/kg and 2.9±1.49N•s/kg, respectively; both p=0.03). Maximum vertical force and loading rate were significantly reduced in the prosthetic limb when compared to the control and intact limbs (83.7±9.3N/kg/ 17.5±2.3N/kg/s vs. 96.8±4.1N/kg/ 25.4±4.4 N/kg/s and 91.2±7.1N/kg/ 23.4±6.7N/kg/s, respectively; all p<0.05), though no significant differences were observed between the intact and control limbs (p=0.18 and p=0.40, respectively) which would appear to be a possible protective effect in the DTTA when compared to the TTA population with regards to osteoarthritis. Limitations of this study are recognized by the population size as well as the heterogeneity in participants with regards to age and varying physical fitness levels. Implications of the current findings support gluteal contribution from the prosthetic limb and indicate a desire for increased stability when walking.

4.1 Steady-state walking velocity

The SSWV in the DTTA was found to be significantly reduced when compared to controls (1.07±0.2m/s and 1.30±0.2m/s, respectively; p=0.03). These current values in both controls and DTTA groups corroborate with the existing literature [26,28,29]. As well, the results of the current study confirm prior findings that DTTA have a reduced SSWV when compared to healthy adults [30,31]. Finally, the result that SSWV was unchanged regardless of leading limb, has also been confirmed in TTA gait initiation [21,32,33]. Previous studies have hypothesized that this invariant SSWV, regardless of leading limb, may be due solely to the strategy used to produce SSWV. Vrieling and colleagues confirmed that the intact limb compensated for the prosthetic limb by increasing propulsion either as leading or trailing limb [21]. Michel and colleagues reported that the anticipatory postural adjustments made prior to gait initiation were different beneath the intact or prosthetic trailing limb. That is, the time taken to initiate gait in the non-preferred condition (i.e. intact leading limb) was greater in order to create increased velocity [32,33]. The APA results related to this study, published in a prior study, outline this time difference. Thus, though the first step kinetics were significantly reduced in terms of propulsion in the prosthetic limb, the overall SSWV achieved at step 4 is invariant, regardless of leading limb.

4.2 Ground reaction forces and impulses

4.2.1 Braking and propulsive impulses

The propulsive impulse demonstrated a significant difference between the prosthetic and both the control and intact limbs (2.0±0.9N•s/kg vs. 3.9±0.6 N•s/kg and 2.9±1.4 N•s/kg, respectively; both $p=0.03$). No significant differences were observed between the control and intact limb ($p=0.18$). For braking impulse, no significant differences were observed between the control, intact or prosthetic limbs (-1.0±0.8 N•s/kg, -1.4±0.8 N•s/kg and -0.8±0.5 N•s/kg, respectively; all $p>0.05$).

Braking and propulsive impulses are important parameters as they comprise both force and time into their calculation. Impulse has been considered in the TTA when walking and measuring this parameter in the TTA has proven valuable as TTA modulate both time and force parameters when walking. Therefore, impulse is perhaps better able to testify of the differences present between the intact and prosthetic limbs when compared to GRF values [39]. Indeed, impulse allows for important insight into how force is modulated [40].

Perhaps the most important result obtained in the current study is that of the propulsive impulse. Indeed, the propulsive impulse in the prosthetic limb was significantly reduced when compared to the intact and control limbs, corroborating prior study [38]. Interestingly, the propulsive impulse in the prosthetic limb is quite large when compared to this previous study. Undeniably, propulsion in the prosthetic limb does not take place through the usual ankle joint propulsors (i.e. gastrocnemius-soleus muscle complex) as they are absent. Compliance from the passive prosthetic foot is perhaps in part responsible for this propulsive impulse [41]. It is theorized that the DTTA utilize the gluteal muscles of the prosthetic limb to propulse during this first step in gait initiation. The large intact muscle would perhaps help in push-off of the prosthetic limb with an above normal level power generation, actively helping the prosthetic limb into swing phase and into the next step, thus compensating for the loss of power from the missing gastrocnemius-soleus muscle complex [10].

When the propulsive impulse results are compared between the intact and control limb, no significant differences were observed. These findings corroborate with prior study [40].

No significant differences were observed across all three conditions for braking impulse. Though previous studies have observed significantly greater braking impulse in the intact limb when compared to the prosthetic limb as well as greater braking impulse in the intact limb when compared to the control limb [16,19,33]. The current results, though means were found to be in this direction, failed to show statistical differences. Again, the reduced values due to the gait initiation task being observed and the heterogeneity of participants are perhaps cause for this.

As considered with regards to increased propulsive and reduced braking forces, greater propulsive force and smaller braking force is seen in the control limb when compared to the intact limb, and moreover when compared to the prosthetic limb. This in large part explains the reduced SSWV obtained in the DTTA when compared to controls, as braking is increased and propulsion is decreased, as reported in previous literature [42,43].

4.2.2 Maximum braking and propulsive forces

The maximum braking force in the prosthetic limb was found to be reduced when compared to the intact limb (-0.3±0.2N/kg and -0.6±0.2N/kg, respectively; p=0.03). No differences were observed between the control and intact limbs (-0.5±0.3N/kg and -0.6±0.2N/kg, respectively; p=0.87) nor between the prosthetic and control limbs (-0.3±0.2N/kg and -0.5±0.3N/kg, respectively; p=0.13). No significant differences were observed between the control, intact or prosthetic limbs for maximum propulsive force (1.5±0.4N/kg, 0.8±0.3N/kg and 0.7±0.4N/kg, respectively; all p>0.05).

To achieve SSWV from quiet standing, increased propulsive and reduced braking force is required [34]. This is seen during the first step of gait initiation across all groups and conditions (control, intact & prosthetic). The significantly increased maximum braking force observed in the intact limb (-0.6 N/kg) when compared to the prosthetic limb (-0.3 N/kg) corroborates prior findings [35,36,37,38,47]. Maximum braking force was not significantly greater in the intact limb when compared to the control limb (-0.5 N/kg) as prior studies have shown [35,36,37]. This is discussed further below in limitations, but these results not being statistically different is probably dependent on the large standard deviation of values obtained due to the heterogeneity of participants.

No maximum propulsive forces were significantly different, though previous studies have shown increased maximum propulsive force in the control limb when compared to the intact limb as well as significantly greater maximum propulsive force in the intact limb when compared to the prosthetic limb [35,36,37].

Additionally, the current results in maximum braking and propulsive force are for the first step in gait initiation. Prior studies have investigated these forces during SSWV [35,36,37]. Thus, the magnitude of these forces is therefore smaller in this first step of gait initiation as SSWV has not yet been reached. It is plausible that these smaller values, and therefore smaller differences in values, hence failed to show significant differences.

4.2.3 Maximum vertical force and loading rate

The maximum vertical force and loading rate were significantly reduced in the prosthetic limb when compared to the control and intact limbs (83.7±9.3N/kg/ 17.5±2.3N/kg/s vs. 96.8±4.1N/kg/ 25.4±4.4 N/kg/s and 91.2±7.1N/kg/ 23.4±6.7N/kg/s, respectively; $p=0.03$/$p=0.01$ and $p=0.04$/$p=0.02$, respectively), though no significant differences were observed between the intact and control limbs ($p=0.18$ and $p=0.40$, respectively).

The maximum vertical force observed in the prosthetic limb was significantly less than the intact and control limbs. This corroborates with previous literature [44]. Surprisingly, no significant difference was observed between the maximum vertical force between the intact and control limbs. Previous studies have observed a significantly increased maximum vertical force in the intact limb when compared to controls [44].

Maximum vertical force is an additionally important biomechanical parameter to measure in the DTTA as studies have indicated that the TTA are at risk of developing compounding complications in their intact limb. Because of the added use of the intact limb in preferred weight bearing, for example, there are important strength and muscle mass discrepancies between the intact and prosthetic limb [44]. Indeed, the intact limb has been shown to be more susceptible to developing osteoarthritis because of the increased demand placed on it when compared to the prosthetic limb [45]. By understanding the mechanisms linked to this increased demand and improving rehabilitation training, there is hope for reduced risk of osteoarthritis in the intact limb [46].

With regards to gait initiation, the DTTA would appear to not be at increased risk of osteoarthritis when maximum vertical force is considered. Though weight bearing is undeniably greater on the intact limb when compared to the prosthetic limb, in the specific situation of gait initiation, there does not appear to be an added risk of osteoarthritis as the values obtained in the intact limb were not significantly greater than those in controls. It is posited that this is due to the significantly decreased SSWV that has been documented in the DTTA when compared to the traumatic TTA. Indeed, by applying a reduced maximum vertical force on the intact limb during gait initiation, the DTTA are perhaps protecting this intact limb from large, detrimental forces which could lead to increased osteoarthritis risk. Conversely, the DTTA, when compared to the traumatic TTA, often lack greater muscular and sensory afferent information in the intact limb [23] and are perhaps then unable to adjust and increase the demand to the intact limb. Studies are warranted in investigating the osteoarthritis risk in the DTTA when compared to their traumatic TTA counterpart.

The loading rate, as with impulse, takes into consideration both force and time, a perhaps more suitable parameter in the study of the DTTA population, as stated above. The results obtained for the loading rate during the first step of gait initiation are in line with those obtained for maximum vertical force. Indeed, loading rate in the prosthetic limb was significantly less when compared to controls and the intact limb. These findings again corroborate with previous literature [44]. The loading rate in the intact limb was not significantly greater than the control limb though previous results have observed this [44]. Again, this will be discussed with regards to participant heterogeneity.

4.3 Gait initiation mechanisms

The results outlined by this study indicate an important difference in the gait initiation mechanisms employed by DTTA and their age-matched controls. Moreover, in the DTTA group, results indicate a significant difference between the prosthetic and intact limb. All DTTA in the current study reported they preferred to initiate gait with their prosthetic limb, employing different strategies in the intact and prosthetic limbs as outlined by the results.

The TTA gait initiation pattern has been described as careful when compared to able-bodied [18]. As stated, gait initiation poses important challenges to TTA, as both balance and propulsion are needed. In order to control for this precarious situation, TTA adopt strategies to favor balance over propulsion, producing a slower, more careful gait. The results of the present study corroborate this notion, as well the DTTA preferring more stable gait initiation by leading with their prosthetic limb[19,21].

As well, in TTA gait, moreover in gait initiation, there is a complex compromise between power generation (i.e. to create greater torque and force to increase SSWV) and balance (i.e. avoiding falls) [22]. Indeed, all DTTA subjects had passive prosthetic feet in the current study. When considering powered prosthetics, passive prosthetics optimizes balance whilst compromising power generation. Implications for future research with regards to this are outlined in the following section.

4.4 Limits, clinical significance, implications & future work

Limitations of this study are recognized by the population size as well as the heterogeneity in participants with regards to age and varying physical fitness levels. The DTTA participants in this study are perhaps more physically fit than the average DTTA population based on the inclusion/ exclusion criteria established and the motivation to participate. As well, the total number of participants included in each test group could perhaps undermine statistical power. Additionally, because of the chosen study population and inclusion criteria, a wide variability with regards to age as well as functionality following amputation (i.e. dysvascularity) in the TTA group were observed. There was important variability observed within each group, between individuals, as can be observed by the large standard deviation values.

Another limitation to this study lies in the choice to average the right and left limb values obtained in the control subjects. Though no statistical differences were found between the right and left limbs, and though the mean and standard deviation values were observed to ensure no outliers, it is possible that future studies comparing right and left limbs distinctly in controls when compared to TTA would allow not only to observe differences in the TTA, but also be able to report on dominant limb and the role of the dominant limb in gait initiation.

For those parameters significantly different, with regards to effect size, large clinical significance was observed for SSWV between controls and DTTA (d=1.15), for maximum braking force between the intact and prosthetic limbs (d =1.50), for propulsive impulse between control and prosthetic limbs (d =3.17), for maximum vertical force between control and prosthetic limbs (d=3.20) and between intact and prosthetic limbs (d=1.06) and finally, for loading rate between control and prosthetic limbs (d=1.80) and between intact and prosthetic limbs (d=0.88). Along with the significant differences observed in these parameters in the above specified conditions, these results of large clinical significance add to the differences demonstrated in the control, intact and prosthetic limbs between the control and DTTA subjects during gait initiation. Thus, these differences observed in the DTTA when compared to healthy adults bear witness of the important strategy difference in first step gait initiation.

As well, for some parameters, large clinical significance was observed, though significant differences were not observed: for maximum propulsive force between the control and intact limbs (d=1.75) and between the control and prosthetic limbs (d=2.00), for propulsive impulse between the control and intact limbs (d=1.67) and finally, for maximum vertical force between the control and intact limbs (d=1.37). Though the clinical significance is sizeable and far exceeds the 0.8 threshold for large clinical significance, conclusions cannot be made because no significant differences were observed for these parameters in these conditions. As mentioned, large standard deviations may be to blame for this as the statistically small number of participants (n=10) and the heterogeneity of participants is notable. Thus, further work is needed into investigating these parameters in the specific DTTA population when compared to controls in order to fully understand the gait initiation strategy employed.

Future studies should aim to compare DTTA and traumatic TTA during a gait initiation task and study into gait termination is also necessary. Also, further work in the aim of creating powered prosthetics is warranted in order to counter for the lack of ankle plantarflexors, all the while reducing compromise with regards to balance control, allowing safe gait, and gait initiation, in TTA and moreover, the DTTA.

Recommendations to rehabilitation training should include intensive and focused practice on complex motor skills such as gait initiation in the DTTA, focus being placed on practice of gait initiation with both the prosthetic and intact limbs to better equip the DTTA population in facing everyday situations, in the hopes of improving function and quality of life in this population.

5. Conclusion

Current results confirm that the mechanisms employed by DTTA in gait initiation to reach SSWV do differ from those employed by able-bodied individuals when initiating gait. Importantly, the propulsive impulse created by the prosthetic limb make plausible prosthetic limb gluteal muscle participation in presence of the missing ankle joint propulsors. The maximal vertical force and loading rate show evidence of a protective factor in the DTTA intact limb, as these do not appear to be greater when compared to controls as previously seen in the TTA.

These altered gait strategies and mechanisms indicate a desire for increased stability when walking, compromising propulsion and velocity of gait. The implications of these findings could translate to the rehabilitation setting with regards to gait initiation relearning. Gait initiation leading with both limbs should be an avid focus in DTTA rehabilitation, as this complex motor task is a critical component of everyday functioning and locomotion.

Conflicts of interest statement

Authors state that no conflicts of interest are present in the research.

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Tables

Table 1- Kinematic and kinetic values obtained in controls and DTTA: prosthetic and intact limb leading. Kinetic values are those obtained during the 1^{st} step in gait initiation. Symbols (*, †) indicate significant differences between groups ($p < 0.05$).

Leading limb:	**Control**	**Intact limb**	**Prosthetic limb**
Steady-state walking velocity (m/s)	$\bar{x}$ = 1.30 ± 0.2* M= 1.28 (0.97: 1.97)	$\bar{x}$ = 1.07 ± 0.2* M= 1.03 (0.56: 1.48)	
Maximum braking force (N/ kg)	$\bar{x}$ = -0.5 ± 0.3 M= -0.5 (-1.0: -0.2)	$\bar{x}$ = -0.6 ± 0.2* M=-0.5 (-1.0 :-0.3)	$\bar{x}$ = -0.3 ± 0.2* M= -0.3 (-0.5: -0.1)
Maximum propulsive force (N/ kg)	$\bar{x}$ = 1.5 ± 0.4 M= 1.5 (0.7: 2.0)	$\bar{x}$ = 0.8 ± 0.3 M= 0.8 (0.4: 1.3)	$\bar{x}$ = 0.7 ± 0.4 M= 0.7 (0.3: 1.4)
Braking impulse (N•s/ kg)	$\bar{x}$ = -1.0 ± 0.8 M= -0.8 (-2.1: -0.01)	$\bar{x}$ = -1.4 ± 0.8 M= -1.2 (-2.7: -0.2)	$\bar{x}$ = -0.8 ± 0.5 M= -0.8 (-1.3: -0.3)
Propulsive impulse (N•s/ kg)	$\bar{x}$ = 3.9 ± 0.6* M= 3.9 (3.1: 4.9)	$\bar{x}$ = 2.9 ± 1.4† M= 3.2 (0.6: 4.5)	$\bar{x}$ = 2.0 ± 0.9*† M= 2.1 (0.7: 3.2)
Maximum vertical force (N/ kg)	$\bar{x}$ = 96.8 ± 4.1* M= 97.9 (89.4: 101.2)	$\bar{x}$ = 91.2 ± 7.1† M= 90.0 (84.7: 102.4)	$\bar{x}$ = 83.7 ± 9.3*† M= 87.9 (73.8: 97.2)
Loading Rate (N/ kg/ s)	$\bar{x}$ = 25.4 ± 4.4* M= 24.2 (19.7: 34.1)	$\bar{x}$ = 23.4 ± 6.7† M= 21.4 (13.1: 36.8)	$\bar{x}$ = 17.5 ± 2.3*† M= 18.9 (13.3: 19.6)

Figures

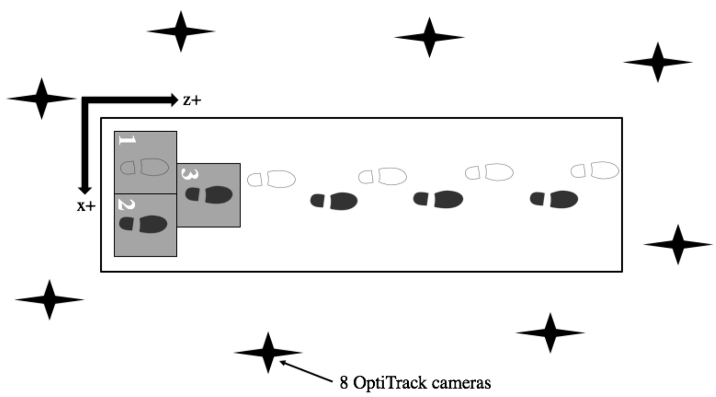

Figure 1- Sketch of experimental set-up. Participants started in quiet standing position, which both feet side-by-side, each foot on one force platform.

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Chapter 6: **Discussion**

The current thesis chose to explore three principal areas of research within gait analysis in healthy adults and the DTTA populations. Because of the very large number of different parameters proposed in the scientific literature that can be measured in healthy adult locomotion, The first purpose of this thesis was to determine the most relevant biomechanical parameters used for gait analysis in a healthy adult population. Thus, a systematic review was carried out and the results indicate that spatio-temporal parameters were those parameters most often measured and walking velocity, cadence and step/stride length were the most often assessed during gait analysis in healthy adults.

The second purpose of the current thesis was to compare the APA's during gait initiation used by DTTA with age-matched controls. It was hypothesized that the APA's used by the DTTA would be altered when compared to healthy controls, more specifically with regards to a reduced posterior CoP_{net} displacement beneath the prosthetic trailing limb in the dysvascular transtibial amputee when compared to the healthy control. The results partially corroborate the hypothesis. Indeed, a small anterior CoP_{net} displacement, rather than a reduced posterior CoP_{net} displacement, was observed beneath the prosthetic trailing limb for APA3 phase in the specific DTTA population when compared to age-matched controls. It is theorized that this anterior CoP_{net} displacement observed beneath the prosthetic trailing limb would be a specific adaptation mechanism in the DTTA when compared to the traumatic TTA in order to improve stability, though reducing propulsion, in the precarious situation of gait initiation.

Finally, the third purpose of this thesis was to compare the underlying biomechanical differences in the gait initiation parameters of DTTA with those of healthy age-matched controls. It was hypothesized that the DTTA would show reduced braking, propulsive and vertical forces during the first step of gait initiation when compared to the healthy controls. As well, it was hypothesized that, in the DTTA, the intact limb would show significantly greater propulsion forces when compared to the prosthetic limb. The results partially support the hypotheses put forward. Reduced braking force, propulsive impulse, vertical force at weight acceptance and loading rate were indeed observed in the prosthetic limb when compared to the intact limb in the DTTA, but no significant differences were observed between the intact and control limbs. Thus, it is put forward that the reduction in SSWV would indeed be a protective factor in terms of intact limb compounding factors in the specific population of the DTTA. Again, these altered gait strategies and mechanisms indicate a desire for increased stability when walking, compromising propulsion and velocity of gait.

Through the systematic review of healthy adult gait, this thesis proposes key biomechanical parameters that are the most relevant for gait analysis. As well, this thesis identifies specific biomechanical behaviours in compensation for the loss of a limb during the anticipatory phases of gait initiation. Finally, specific mechanisms were identified in the DTTA population to accelerate the CoM and reach SSWV. The role of this general discussion is to explore the results and themes presented in the three separate gait analysis studies carried out in the current thesis. As well, the work carried out investigates how DTTA modified the biomechanics of their gait pattern compared to healthy adults. Because of the disease related problems seen in DTTA, understanding gait in this specific population needs to be undertaken distinct from the TTA for other causes. Perhaps in understanding the unique biomechanics, strategies and mechanisms used in the DTTA population when compared to the healthy adults, health professionals would be better equipped in providing rehabilitation programs and prosthetic fitting to accommodate the needs of this specific DTTA population.

This general discussion will first discuss the most relevant parameters for gait analysis will be determined with a systematic review of the existing literature. Then, the APA's, prior to and during gait initiation, as well as the kinetics during the first step of gait initiation in order to increase the walking velocity to reach SSWV in the DTTA when compared to healthy adults will then be explored.

1. Most frequently measured biomechanical parameters in healthy adults.

A note would first like to be made with regards to the term *most relevant* used in the first paper of the current thesis. After further reflection, we find that the term *most relevant* to be perhaps incorrect as many other factors, discussed further below, interplay in this decision and the term *most frequently reported* to perhaps be more exact in this systematic review then. This paper has since been published and therefore changes to the paper will not be made to conserve the integrity of the work. However, we find that the term *most frequently measured* to be a more appropriate term at this time, and thus, this term will be used hereon after.

Indeed, the term most relevant should be patient oriented. That is, most relevant should be able to discriminate for improvements or any irregularities in healthy adults gait. At this time, we are uncertain that the parameters which were most frequently measured in the selected articles can also discriminate between improvements in quality of life in the healthy adults population.

Results obtained by this systematic review of 65 articles of gait analysis in healthy adults were first tabulated among single parameters as well as the number of articles measuring this given parameter. Then a summation of parameters was carried out to group sub-parameters with a given parameter. This summation of parameters is important, as also carried out by Sagawa et al., as often such parameters as ankle force can be divided among many various sub-parameters (i.e. minima and maxima values, various planes, phases of gait cycle, etc.), therefore inflating the number of times this parameter was measured. For example, though power, work and energy type of parameters were measured most often, no single power, work or energy parameter was reported more than 10 times and most were only measured once. Results with regards to angle, moment, power, etc. at the joints may therefore be inflated as often various sub-parameters are derived from a same measurement. Thus, the spatio-temporal parameters, namely walking velocity, cadence, stride length and step length were those single parameters most often measured and will be discussed in light of the calculations carried out below.

The Relevance score is perhaps the most important calculation carried out. Indeed, by creating a score which accounts for both frequency of measurement and number of articles measuring a given parameter, objectivity to this tabulation of parameters is added. Walking velocity was observed to be the most frequently measured parameter, followed by stride length (0.879 and 0.686, respectively). Further systematic reviews should aim at developing these scores to better quantify to the relevance of parameters.

The mean attributed Level of Evidence score was found to be high (11.8±1.8 out of 14 possible points). As well, the reduced number of articles included in this systematic review in healthy adults when compared to that of Sagawa and colleagues [105], lead to believe that the Evidence score was not discriminatory enough in its scoring or perhaps that inclusion and exclusion criteria weeded out the lower quality articles. A Level of Evidence score with a wider array of possible scores is needed.

With regards to both correlations carried out, between frequency of parameters and Level of Evidence (r_s =-0.224, p=0.06) and between Level of Evidence and Journal Impact Factor, no relations were found (r_s=-0.133, p=0.105). The absence of relation between Level of Evidence and Journal Impact Factor was as well found by Sagawa et al. [105].

The results of this systematic review corroborate some of the parameters selected for the lower limb amputee. Walking velocity, cadence and step/stride length appearing to be the most frequently measured biomechanical parameters to healthy adult gait analysis. Walking velocity is an encompassing parameter, as mentioned by Sagawa and colleagues. Also, such parameters as cadence, stride length, etc. are components to the calculation of SSWV and therefore affirm the importance of SSWV as a key parameter for gait analysis. Along with ease of measurement and cost efficiency, walking velocity and other spatio-temporal parameters would therefore appear to be the most frequently measured biomechanical parameters for gait analysis in healthy adults.

The systematic review conducted by Sagawa et al. sought to identify the most relevant biomechanical and physiological parameters for assessing gait in individuals with different levels of lower limb amputations. Walking velocity and associated parameters, joint angular position of the lower limb articulations and kinetics recorded from force platforms, were those parameters most often measured in the articles included. As well, the ease of measurement with which the spatiotemporal parameters can be measured was of particular importance. Unfortunately, due to lack of overall quality of articles included and the parameter diversity in the lower limb amputee gait analysis, Sagawa et al. warrants that further research is needed.

To our knowledge, this is a first systematic review of its kind in a healthy adult population and the implications of these findings are important for choosing the most relevant biomechanical parameters for gait analysis. Further work should be carried out to establish relevance of parameters in light of the author's expertise as well as the use of various equipment. Future studies should also aim to identify if the most relevant biomechanical parameters for gait analysis found in healthy adults are also relevant to other clinical populations. Individuals with a transtibial amputation and healthy adults yielded the same parameters, but perhaps the results obtained in other populations would be different, such as in populations with a neurological disorder (i.e.: Parkinson's, Stroke or Cerebral Palsy) or with a more severe mechanical impairment (i.e.: hemipelvectomy amputation).

2. APA's for gait initiation in the DTTA

Perhaps the most important result in the investigation of the APA pattern in the DTTA is with regards to the APA_{total} A/P CoP_{net} displacement. An anterior total CoP_{net} displacement was observed in the prosthetic trailing limb condition when compared to a posterior APA_{total} A/P CoP_{net} displacement observed in controls (mean anterior 3.2 cm vs. posterior 3.6 cm, respectively; $p=0.05$). In the healthy adults, and as seen in results of the current thesis, the

APA_{total} A/P CoP_{net} displacement is posterior, in keeping with the notion that this helps in unbalancing the CoM in a forward direction [42]. Previous research in the TTA found a reduced posterior CoP_{net} displacement beneath the prosthetic trailing limb [71,72], while other results were inconclusive with regards to a reduced posterior or anterior CoP_{net} displacement [54,74]. To our knowledge, this thesis was the first to investigate the APA gait initiation strategies employed in solely the DTTA population, and therefore, it is posited that the anterior total APA CoP_{net} displacement would be a strategy employed by the DTTA who show greater physical deterioration compared to the traumatic TTA [28,105].

Indeed, though the average total A/P CoP_{net} displacement was anterior in the trailing prosthetic limb condition, when looking at individual results, two participants displayed a posterior, although reduced, total A/P CoP_{net} displacement rather than an anterior displacement. Based on BMI, age, investigator observations and conversations with participants, these two DTTA individuals were the most physically fit when compared to other participants. Though the DTTA and traumatic TTA have seldom been considered as different entities with regards to gait initiation research, the current results suggest that gait initiation investigation must be specific to cause of amputation in the TTA.

The results with regards to the anterior total CoP_{net} displacement displayed in the trailing prosthetic limb condition in the DTTA therefore imply that the inverted pendulum strategy employed in controls, in pushing and propulsing the CoM forward, does not work as efficiently in the DTTA when the intact limb is leading (i.e. non-preferred leading limb) [75]. The reduced utilization of the excursion of the CoP_{net} to produce forward CoM torque in the DTTA leads to reduced propulsion created by the APA towards gait initiation, contributing to the observed reduced gait initiation velocity in DTTA. However, though the strategy employed by the DTTA reduces propulsion, a more stable balance state in gait initiation is possible [54,70,74].

As well, results showed no significant differences between controls, the prosthetic and the intact limbs with regards to the duration of APA1 phase (0.49 s, 0.58 s & 0.56 s, respectively; all $p > 0.05$). No results displayed significant differences between controls, prosthetic and intact limb for A/P CoP_{net} displacement (3.5 cm, 2.2 cm and 2.1 cm, respectively: all $p>0.05$) and M/L CoP_{net} displacements in APA1 (6.4 cm, 3.9 cm and 6.8 cm, respectively; all $p > 0.05$). This is in line with the previous results published on gait initiation in TTA [54,74]. During APA1 phase, in healthy adults, there is a displacement of the CoP_{net} posteriorly and laterally, toward the leading limb. This first phase of APA allows the CoM to shift from a position almost at the center of the base of support during quiet standing toward the leading limb in order to push the CoM forward and to the trailing limb.

With regards to the APA2 phase, the results showed an increased duration time for APA2 phase for both prosthetic (0.39 s) and intact trailing (0.32 s) limbs when compared to the controls (0.19 s) ($p=0.01$ and $p=0.01$, respectively).The APA2 phase is characterized by a displacement of the CoP_{net} towards the trailing limb, as the BW is progressively transferred from the leading limb onto the trailing limb. The end of the second phase occurs when the CoP_{net} is approximately centered between both limbs. This allows for heel-off of the leading limb to occur. There are several reasons to explain this increase in APA2 time in the DTTA. First, it has been reported that the TTA take more time to initiate gait [54,72,74]. Secondly, by increasing the time taken to complete APA2 phase, the DTTA spend more time in double limb stance, a more stable condition. The lengthening of the APA2 phase duration also corroborates with the notion that the DTTA prefer to support their BW on their intact limb. As well, as related by Michel & Chong (2004) findings, the TTA use a different local strategy to produce propulsion impulse. That is, because force produced by the prosthetic limb is significantly diminished, the TTA apply this reduced force for a longer period of time, thus producing greater propulsive impulse [73]. The following paragraph will discuss the increase in the CoP_{net} displacement during APA2 phase.

In controls during APA2, there is a slight anterior shift in CoP_{net} displacement whose aim is to bring the CoP_{net} relatively near the position at quiet standing, as the heel of the leading limb leaves the ground. The results of the current thesis showed a significant increase in APA2 anterior CoP_{net} displacement in the trailing prosthetic limb (5.51 cm) condition when compared to both the intact (2.12 cm) and control (2.31 cm) trailing limb conditions ($p=0.04$ and $p=0.01$, respectively). As the CoM is travelling forward, given the propulsive torque produced by the posterior CoP_{net} displacement in APA1, the strategy adopted by the DTTA to send the CoP_{net} more anteriorly is unique. This has not been observed in other studies of APA in gait initiation of the TTA, probably due to method design [106]. It is theorized that this strategy employed by the DTTA reduces the propulsive effect created during APA1 to produce a more stable condition. That is, the DTTA move the CoP_{net} to a position closer to the CoM which has been projected forward. This supports the theory that the DTTA use a careful gait, prioritizing stability over propulsion [54,70,74].

The APA3 phase also investigated, is characterized by a posterior and lateral CoP_{net}

displacement towards the trailing limb and takes place as leading limb toe-off occurs. Results showed a significant reduction in M/L CoP_{net} displacement during APA3 phase in the intact trailing limb condition (3.59 cm) when compared to the prosthetic (7.11 cm) and control (7.05 cm) trailing limbs (p=0.04 and p=0.05, respectively). During DTTA quiet standing, the CoP_{net} is located closer to their intact limb as weight bearing is increased in the intact limb when compared to the prosthetic limb [48,136]. Thus, in APA2, a position approximately about the quiet standing position, in the DTTA the CoP_{net} position is closer to the intact limb. As well, this is confirmed by the results of displacement of the CoP_{net} during APA1 which are not augmented in the M/L direction, though the CoP_{net} travels laterally to the prosthetic limb. Again, this condition increases stability in the DTTA as the intact limb is advantageously utilized.

Finally, with regards to the explored APA_{total}, for time, A/P and M/L CoP_{net} displacement, from quiet standing to the end of APA3, results indicate a significant increase in APA_{total} time in the prosthetic trailing limb condition when compared to the control trailing limb (1.11s and 0.81s, respectively; p=0.05). This concurs with results of prior studies in which an increase of gait initiation time has been observed in the TTA [54,71,72]. As the APA's are an important component of total gait initiation time (i.e. from start of APA phase to step 3 in gait initiation), and as discussed with regards to APA2 time, it is coherent that the time taken to complete the APA_{total} would also be prolonged [54,71,72]. As well, the strategy used to increase time of application of a reduced propulsive force to create greater impulse in the DTTA warrants that the total APA time would be prolonged [73].

This more stable state, whilst compromising propulsion in gait initiation, makes proof of a 'careful' strategy selection in the DTTA. Prior studies have put forward this notion, the TTA prioritizing stability over propulsion [54,70,74]. This, as well, is observed with regards to the underlying kinetics during the first steps of gait initiation through to SSWV and therefore will be further discussed in the section below.

3. Kinetics contribution of the first step in gait initiation

After the initial APA's, prior to and during gait initiation, the first step is an important component leading to SSWV. As stated, at the first step in gait initiation, approximately 75-90% of the total SSWV is reached [42,53,69,70]. Therefore, the first step in gait initiation is important in producing the remaining forces necessary to reach SSWV. However, though forward movement has been achieved with the APA's in initiating gait, to allow the CoM to continue to travel forward, important propulsive forces must be produced during this first step. Studies have indicated important differences between the population of traumatic TTA versus the DTTA [28,105]. To our knowledge, no literature has yet investigated the kinetic strategies employed during the first step of gait initiation in the specific population of the DTTA while few studies have investigated the first step kinetics in the TTA [54,70,72,74]. Thus, the final objective of the current thesis was to compare the underlying biomechanical differences of the gait initiation parameters employed by the DTTA to those of healthy age-matched controls during the first step of gait initiation.

During the first step of gait initiation, a reduced maximum braking force was observed in the DTTA beneath the prosthetic limb (-0.3 N/kg) when compared to the intact limb (-0.6 N/kg) ($p=0.03$). To increase walking velocity, one can increase propulsive forces or reduce braking forces [107]. Previous research into TTA gait initiation have well documented this during SSWV [66,137,138,139,140]. However, maximum braking force was not significantly greater in the intact (-0.6 N/kg) limb when compared to the control limb (-0.5 N/kg) (p=0.87) as prior studies have shown [66] [108] [138]. This is discussed further below in relation to the protective mechanism in the DTTA.

Results also yielded a significantly reduced propulsive impulse in the DTTA prosthetic limb (2.0 N·s/kg) when compared to both the intact (2.9 N·s/kg) and control (3.9 N·s/kg) limbs during the first step of gait initiation (p=0.03 and p=0.03, respectively). This reduced propulsive impulse is in agreement with prior results in SSWV observed in the TTA and controls [140]. As discussed, the major propulsors in gait initiation are the inverted pendulum mechanism (i.e. CoM and CoP_{net}) and the ankle plantarflexor muscle complex propulsive force and impulse [42,141,142]. Though, the inverted pendulum mechanism produces less forward propulsion during APA's as discussed above, the missing gastrocnemius-soleus muscle complex in DTTA is also an important cause for lack of propulsion during the first step in gait initiation. Interestingly, the intact limb does not produce significantly more propulsive impulse than the control limb, a compensation mechanism that could help in counteracting the reduced propulsion impulse in the prosthetic limb (0.03 N/kg and 0.04 N/kg, respectively: p=0.18). The maximum propulsive force observed between the intact and prosthetic limbs were not significantly different (0.8 N/kg and 0.7 N/kg, respectively: p=0.31). Thus, in order to generate greater propulsive impulse, the time spent applying the propulsive force in the intact limb is therefore increased to create the greater impulse observed. This strategy has been discussed with regards to gait initiation in the TTA and is as well supported by the increased stance time spent on the intact limb when compared to the prosthetic limb [54,72].

As well, though the ankle joint along with its attached muscle components, is missing in the prosthetic limb, the observed propulsive impulse generated is not null. Indeed, some simple compliance of the passive prosthetic material may in part be responsible for this larger than expected propulsive force and impulse in the prosthetic limb, but evidently not solely responsible as lower force values have been observed by prosthetic compliance [100]. It has been theorized that there is a contribution of the gluteal muscles at the hip joint in order to compensate for the deficient gastrocnemius-soleus complex [11]. This gluteal contribution would aid in the forward CoM propulsion while on the prosthetic limb, rather importantly, during the first step of gait initiation. Since DTTA do not have the necessary plantarflexor ankle muscles to create propulsive forces, this energy can be generated by the extensor hip muscles in a "push from behind" strategy as suggested by Winter & Sienko [11].

Maximum vertical force at weight acceptance and loading rate were significantly reduced in the prosthetic limb (83.7 N/kg and 17.5 N/kg/s, respectively) when compared to both intact (91.2 N/kg and 23.4 N/kg/s, respectively: p=0.04 and p=0.02, respectively) and control (96.8 N/kg and 25.4 N/kg/s, respectively: p=0.03 and p=0.01, respectively) limbs during the first step of gait initiation. These results are corroborated with previous literature [109]. Reduced step length [55], velocity of gait initiation [54,70,74] and preferred weight bearing on the intact limb [48,136] are all key reasons for the reduced maximum vertical force and loading rate observed in the prosthetic limb of the DTTA when compared their intact limb and that of age-matched controls.

Interestingly, no significant differences were observed between maximum vertical force at weight acceptance nor loading rate in the intact limb when compared to controls (results displayed above: p=0.20 and p=0.40, respectively). Lloyd and colleagues found a significantly increased maximum vertical force and loading rate in the intact limb when compared to the prosthetic limb in the TTA, related to an increased incidence of osteoarthritis in the intact limb in the TTA [109,110]. However, these studies have not investigated the risk of osteoarthritis in the specific DTTA population. As presented in the introduction, in the DTTA, reduced overall survival rate, postural stability in quiet standing and physical capacity in walking (VO2max) have been reported, when compared to their traumatic counterpart [27,29,30]. It is theorized that this reduction in increased maximum vertical force and weight acceptance and loading rate observed during the first step of gait initiation could be a DTTA-specific protective factor against osteoarthritis in the intact limb. Thus, the careful gait initiation strategy employed by the DTTA to favor stability over propulsion, along with reduced SSWV and overall lower physical fitness capacity could also, inadvertently, aid in protecting against intact limb osteoarthritis.

Participant recruitment was an important limiting factor with regards to the second and third objectives of this thesis. As mentioned in the Methods section, the main outcome parameters (i.e. A_{total}, T_{total}, SSWV, propulsive impulse) established a sample size of 25 participants per group. Alas, over 110 potential DTTA participants were contacted by the recruitment officer of the CRIR and by the first author of the current study over a 1.5 year time period for total participation 10 DTTA for the current study. However, with 2 cohorts of 10 subjects, the results showed significant differences with a clinical significance. Thus, future work is needed to further establish the differences in the DTTA population, leading to better understanding, rehabilitation and quality of life in this distinct population.

Chapter 7: **Conclusion**

In the current thesis, gait initiation, from quiet standing through to SSWV, was explored in the DTTA and in healthy adults. First, a systematic review of the literature was carried out to identify the most relevant biomechanical parameters for gait analysis in healthy adults. Then, this thesis explored the APA pattern and underlying first step kinetics in the DTTA when compared to healthy controls prior to and during gait initiation.

In the systematic review of the literature, spatio-temporal parameters were found to be the most often biomechanical parameters reported by the greatest number of articles. Walking velocity, cadence and step/stride length appearing to be the most frequently measured biomechanical parameters for gait analysis in the healthy adult population. Further research to compare the most relevant parameters for gait analysis in healthy adults and TTA with that of other pathological populations.

The most important result obtained in the APA of DTTA when compared to healthy adults is with regards to the total anterior CoP_{net} displacement observed in the prosthetic limb when compared to a total posterior CoP_{net} displacement observed in healthy controls. These results would appear to be related to further reduction in instability associated with dysvascular amputation, the DTTA favoring stability over propulsion, accounting for the increased APA total time and reduced SSWV observed. It is theorized that the total anterior CoP_{net} displacement would be specific to the DTTA when compared to the traumatic TTA.

Finally, the first step underlying biomechanics reported decreased maximum braking force in the prosthetic limb when compared to the intact limb, though no significant differences were observed with controls. As well, propulsive impulse was significantly reduced in the prosthetic limb when compared to intact and control limbs. These reductions in A/P force/impulse testify of the missing plantarflexor muscles at the prosthetic foot. With regards to vertical forces, maximum vertical force at weight acceptance and loading rate were significantly reduced in the prosthetic limb when compared to both intact and control limbs. Interestingly, no increase in those parameters was observed in the intact limb when compared to the control limb. This can be seen as an osteoarthritic prevention mechanism of the intact limb. Again, these results support the idea of careful gait initiation in the DTTA when compared to controls, confirming that the need for stability is favored over propulsion.

The implications of the current thesis are applicable to the rehabilitation setting. Rehabilitation specialists should focus on prosthetic and intact limb as leading limbs for gait initiation to aid DTTA in everyday perturbations. As well, the current findings are important to the development of powered prosthetic devices, especially for the DTTA who possess further stability reductions when compared to their traumatic counterparts. Future research should focus on comparing gait initiation in the DTTA when compared to the traumatic TTA counterpart.

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