Trunk biomechanics during hemiplegic gait after stroke : a systematic review

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Abstract

Stroke commonly results in trunk impairments that are associated with decreased trunk coordination and limited trunk muscle strength. These impairments often result in biomechanical changes during walking. Additionally, the so-called pelvic step might be influenced by these impairments. Therefore, the aim of this review was twofold. First, to gain more insight into trunk biomechanics during walking in stroke patients compared to healthy individuals. Second, to investigate the influence of walking speed on trunk biomechanics. The search strategy was performed by the PRISMA guidelines and registered in the PROSPERO database (no. CRD42016035797). Databases MEDLINE, Web of Science, Cochrane Library, ScienceDirect, and Rehabdata were systematically searched until December 2016. Sixteen of the 1099 studies met the eligibility criteria and were included in this review. Risk of bias was assessed by the Newcastle-Ottawa Scale. The majority of studies reported on trunk kinematics during walking, data on trunk kinetics and muscle activity is lacking. Following stroke, patients walk with increased mediolateral trunk sway and larger sagittal motion of the lower trunk. Although rotation of the upper trunk is increased, the trunk shows a more in-phase coordination. Acceleration of the trunk diminishes while instability and asymmetry increase as there are less movement towards the paretic side. However, it is of great importance to differentiate between compensatory trunk movements and intrinsic trunk control deficits. Specific exercise programs, assistive devices and orthoses might be of help in controlling these deficits. Importantly, studies suggested that more natural trunk movements were observed when walking speed was increased.

Keywords: Stroke, trunk, biomechanics, kinematics, walking

Trunk biomechanics during hemiplegic gait after stroke: a systematic review

Introduction

During normal walking, the upper and lower trunk move in a coordinated yet opposite direction around the vertical body axis. Trunk motion in healthy individuals is characterized by a flexion peak near each heel strike in the sagittal plane and it reaches maximal range of motion in the frontal plane at the time of toe off [1]. Several studies already concluded that the trunk plays an important role during hemiplegic gait in adult stroke patients [2,3]. Moreover, alterations in trunk kinematics during walking were seen in children with hemiplegic cerebral palsy when compared to typical developing children [4,5]. Therefore, it seems that the trunk significantly alters during pathological gait.

In patients suffering from stroke, trunk impairments are commonly reported. These impairments are characterized by a diminished siting balance, trunk coordination and muscle strength [6,7]. In contrast to the extremities, the trunk is bilaterally impaired. Therefore, both the paretic and non-paretic side of the trunk are characterized by reduced activity levels, delayed onset times, and diminished synchronization of the trunk musculature [8]. Although these impairments are well documented, little is known regarding their influence on trunk biomechanics during walking in stroke patients. Studies have shown that because of these impairments the trunk showed increased mediolateral movements in the frontal plane during sitting, standing and sit-to-stand transfers [9-11]. Since kinematics of the trunk change during various locomotor activities, it seems reasonable that trunk kinematics also change during gait. Therefore, knowledge concerning changes in trunk biomechanics after stroke should be increased so that they can be addressed adequately in therapy.

Additionally, alterations in trunk biomechanics and especially pelvic motion might influence the so-called pelvic step. This phenomenon explains that at a certain walking speed pelvic rotations start to contribute to step length. In healthy individuals, the upper and lower trunk move more out of phase when walking speed increases by changing the timing of the pelvis. However, patients might not use this strategy as they perhaps want to avoid large rotations [12]. On the other hand, larger pelvic rotations might be a compensation for limited hip flexion [13]. Therefore, increasing our knowledge concerning the effect of walking speed on trunk kinematics might redirect gait rehabilitation strategies.

The purpose of this review is to gain more insight into trunk kinematics, trunk kinetics, and muscle activation during walking after stroke compared to healthy individuals. Additionally, the effect of walking speed on trunk biomechanics was investigated to optimize exercise programs.

Methods

Systematic literature search. This review was conducted according to the Preferred Reporting Items for Systematic Review and Meta-Analysis Statement (PRISMA) and registered in the PROSPERO database (no. CRD42016035797). A systematic search of the electronic databases MEDLINE (Pubmed), Web of Science (Web of Knowledge), Cochrane Library, ScienceDirect, and Rehabdata was conducted using the PICO process. The search strategy used were combinations of the following search terms: “cerebrovascular disorders”, “hemiplegia”, “stroke”, “biomechanical phenomena”, “mechanical phenomena”, “kinetics”, “kinematics”, “muscle activity”, “electromyography”, “gait, “walking”, “locomotion”, “mobility”, “Duchenne gait”, “trunk lean”, “Trendelenburg gait”, “trunk”, “torso”, “pelvis” and/or “thorax”. The specific search strategy per electronic database is provided in appendix A. The final systematic literature search was performed in December 2016.

Eligibility criteria. The studies had to meet the following inclusion criteria: 1) adult participants (age 18 or older) with primary diagnosis of stroke and a control group of healthy individuals. No control group was necessary when the effect of walking speed was investigated, the same stroke population had to be investigated during several walking speed conditions; 2) outcome measures assessing kinetic and/or kinematic parameters of the trunk, and/or electromyographic (EMG) activity of the trunk muscles; 3) outcome measures had to be assessed during walking, specifically during a basic walk across. Studies examining walking actions such as sit-to-stand, turning or obstacles crossing were excluded from this review; and 4) all designs, except systematic reviews, meta-analysis and case studies were included in the study. Furthermore, studies investigating the effect of a certain intervention program or were not written in English, Dutch, German or French were excluded. No limitations were applied regarding the time after stroke onset or the publication dates of the included studies.

Outcome measures. To have a good understanding of the primary and secondary outcome measures, it is important that the trunk is well defined. The trunk consists of the following four segments: the shoulder girdle, thorax, abdomen, and pelvic girdle. The shoulder girdle consists of the scapula and clavicle, the thorax is located between the neck and abdomen, the abdomen is located between the thorax and the pelvis, and the pelvic girdle consist of the pelvis, pelvic cavity, and sacrum. In the current study, the shoulder girdle and thorax was defined as the upper trunk (UT), in contrast to the lower trunk (LT) consisting of the abdomen and pelvic girdle. The following three parameters were considered primary outcome measures: 1) kinematics which describes the displacements and/or range of motion (ROM) of the trunk in the sagittal, frontal, transversal plane. The displacement with respect to time also known as angular and linear velocity, and the change in velocity with respect to time known as angular and linear acceleration. Data were collected by means of a full body gait analysis or tri-axial accelerometers; 2) kinetics which examines the forces acting on the body during movement, measured by means of force plates or calculated through inverse dynamical analysis; 3) muscle activity of the abdominal and back muscles which is assessed by means of electromyography, providing information on the timing and amplitude of muscle contractions.

Study selection. Three reviewers (SV, PJV, and TVC) independently screened for eligibility based on title and abstract. The remaining studies were independently assessed based on full-text evaluation. Subsequently, reference lists were hand searched for additional studies not yet obtained during the systematic search. Consensus was sought between the three reviewers when disagreement occurred.

Risk of bias. To assess the methodological quality of non-randomized controlled trials, the Newcastle-Ottawa Scale (NOS) was used. According to the design of the study, the checklist for case-control, cohort, or cross-sectional studies was employed. The NOS assesses the risk of bias by means of a star rating system, all studies are judged on three categories: selection, comparability, and exposure or outcome. A star can be awarded when a predefined criterion is met, with a maximum of nine stars to be obtained. Rewarding a predefined criterion with a star suggests that this criterion has a low risk of bias. Within the category comparability and the subcategories ascertainment of the exposure risk and assessment of outcome a maximum of two stars can be given so a clear hierarchy in response is present. No clear cut-off values are known for the NOS, therefore values described in McPheeters et al [14] were used. A score of 7 or higher was considered good, a score between five and seven was moderate, and lower than five was defined as poor. The developers of the NOS have established the face and criterion validity, and inter-rater reliability. Concerning cross-sectional studies, the adapted version of the NOS, by Herzog et al [15] was used. Two independent researchers assessed the risk of bias. A third researcher was consulted when disagreements occurred. At last, the level of evidence of each study was determined by means of the Grading of Recommendations Assessment, Development and Evaluation (GRADE) approach. A score of A1 is given to systematic reviews, A2 to randomized double-blinded controlled trials, B to randomized clinical trials or comparative studies of moderate quality, and C to non-comparative studies.

Data extraction. Data were extracted by the three reviewers and summarized in a methodology and evidence table (Table 1 and 2). Study design, study population, gait analysis equipment, study protocol, and outcome measures were outlined in the methodology table.

The first outcome measure, trunk kinematics, was subdivided into three categories: motion of the trunk and its segments, motion of the segments relative to each other, and the variability of trunk motion. These three subdivisions of trunk kinematics were discussed in the evidence table and defined as:

First, linear and angular trunk kinematics in all planes described the movements of the upper, lower, and whole trunk. Both range of motion (ROM) as well as accelerations were included in this category.

Second, trunk coordination described the movements of the upper and lower trunk relative to each other. An important distinction has to be made between arm-leg coordination. Trunk coordination was defined as the continuous relative phase (CRP) of the shoulder and pelvic girdle which describes the difference in both position and velocity between the two body segments. A phase difference of 180° represents a perfect antiphase movement compared to a complete in-phase movement which has a phase difference of 0°. An antiphase movement, as seen in normal walking, suggests that shoulder and pelvic girdle move in opposite direction.

At last, trunk variability described the variance of trunk movements and was mostly described as trunk symmetry and trunk stability. Trunk symmetry was assessed by determining the symmetry ratio of the paretic and non-paretic side and asymmetry index or symmetry score which is quantified relative to a calculated midpoint between the feet. A symmetry score or asymmetry index of zero indicates symmetry between both sides. The magnitude of asymmetry describes the degree of asymmetry and the direction of asymmetry is represented by the sign given to the index/score. A large positive or negative number indicates more asymmetry towards the paretic or non-paretic side respectively. Subsequently, trunk stability was defined by the following three parameters: 1) the root mean square (RMS) is a measure of dispersion and can be used as mean RMS which is computed by averaging four RMS values related to four analysed strides and normalized RMS which indicates the magnitude of trunk fluctuation without being influenced by walking speed. Techniques to determine RMS are well established in gait analysis [16]; 2) local stability can be described as the response to perturbations in real time and quantifies how much each trajectory moves towards or away from its own nearest neighbour trajectory. Local instability can be quantified by computing the short-term local divergence exponent (LDE); 3) orbital stability is a response to perturbations from one cycle to the next and quantifies how much each trajectory moves towards or away from the central cycle trajectory and was estimated by the maximum Floquet multipliers (maxFM). Both local and orbital stability have been thoroughly described in the study of Dingwell et al [17].

The second and third outcome measure, trunk kinetics and muscle activity was not described by any of the included studies. Therefore, we will not elaborate it.

At last, the effects of walking speed on trunk kinematics described which of the previous parameters are speed-dependent.

Results

Search results. The search strategy resulted in 1099 eligible studies obtained from electronic databases MEDLINE (Pubmed), Web of Science (Web of Knowledge), Cochrane Library, ScienceDirect, and Rehabdata. After deduplication, 165 studies were potentially relevant for a more detailed screening based on title and abstract. After full-text reading, 152 papers did not meet eligibility criteria and were excluded. In addition, three studies were retrieved by hand searching of the reference lists of the included studies. A total of 16 were identified as relevant and were included in this review (Figure 1).

Risk of bias. Methodological quality was assessed using the Newcastle-Ottawa scale for quality assessment of non-randomized trials as seen in Table 1a and 1b. Thirteen case-control studies were included, the remaining three articles were cross-sectional studies. In general, the median score for the case-control studies was six out of nine and for the cross-sectional studies six out of ten, indicating a moderate risk of bias. The two researchers had an 82% agreement on the scales, a third researcher decided when no consensus was found. Following the GRADE approach, Thirteen studies had a level of evidence B which categorizes the study as a comparative study. The remaining three studies are non-comparative studies and are therefore classified as having a C-level of evidence.

Study characteristics. Study characteristics are shown in Table 2. The examined outcome measures were trunk kinematics and muscle activity. None of the studies examined trunk kinetics during walking in people suffering from stroke. The different devices used for gait analysis to examine trunk kinematics can be categorized into the following groups: 1) 3D motion capture [18-29]; 2) tri-axial accelerometer [30-32]; and 3) 2D video [33]. Sampling rate varied from 25 to 200Hz in the included studies. In the studies using accelerometers, devices were placed on the L3 spinous process. Reflective markers used during 3D motion capture were placed on various anatomical landmarks. Upper trunk motion was assessed by placing markers on C7, left and right acromion, and/or mid-sternum. For the lower trunk markers were placed on the left and right posterior superior iliac spines (PSIS), S2 spinous process, mid-point between PSIS, or pelvis.

Furthermore, parameters were measured during over ground walking on a 7, 10, 15, or 16 meter walk way [23,25,26,28-32], or during treadmill walking [18-22,24,27,33]. Patients were asked to walk at their self-selected speed in fourteen studies [18-26,28-32], in six studies patients were also asked to walk at their maximal speed [19-21,24,32,33], and in five studies at a percentage of their maximal or self-selected walking speed [19,22,24,27,33]. The majority of studies specified that patients wore their own shoes during the dynamic walking trials [22-24,28,30-32]. One study reported that patients were obligated to walk barefoot [26]. Four studies mentioned that patients could walk with an orthosis if necessary [24,28,31,32]. In two studies, patients were allowed to walk with walking aids, more specifically a cane [30,31]. Seven studies specified that patients were not allowed to walk with any walking aid or support of the treadmill [18,20,26-29,32]. Six studies did not mention the walking conditions [19,21-23,25,33].

The majority of the studies examined trunk kinematics [18-20,22,30-33]. In addition, lower trunk displacements were described by six studies [23,25,26,28,29,33]. Others investigated the effect of walking speed on trunk kinematics [19-24,27,30,33].

Mean time post stroke varied from one month to six years after stroke diagnosis. Nine studies included chronic stroke patients with a mean time post stroke of two to six years [18-22,24,26,28,31]. Four studies included sub-acute stroke patients with a mean time post stroke of one to six months [23,25,30,32]. Three studies did not mention time post stroke [27,29,33].

Synthesis of results.

Linear and angular trunk kinematics. Eight studies examined the displacements of the upper, lower, or whole trunk [20,22,23,25,26,28,29,33].

In the frontal plane, two studies suggested enlarged lateral sway of the lower and whole trunk which were oriented towards the non-paretic side during stance phase [20,23]. In addition, trunk accelerations of the upper and lower trunk were significantly larger in stroke patients as in controls [20]. Significant differences were found in the pelvic obliquity angle and velocity at toe off [28] and during swing phase [29]. Stroke patients showed increased motion and a upward pelvic movement during the swing phase, which is in contrast with the downward movement seen in healthy individuals.

In the sagittal plane, two studies investigated upward and downward pelvic tilt and concluded that stroke patients showed excessive anterior pelvic tilts (> 4 cm) during the stance and swing phases [25,26]. In contrast to the good stroke group in the stroke population (Brunsnström stages 5-6/near normal gait) and the healthy subjects, the poor stroke group (Brunnström stages 3-4/decreased knee flexion) tilted their pelvis on the paretic side upward (2.18 $\pm$ 3.77 cm) during stance phase and downward (-6.6 $\pm$ 4.7 cm) during swing phase [25]. Normally, a downward movement is seen during the stance phase and an upward movement during the swing phase equal to or less than three cm [25,26].

In the transverse plane, ROM in the upper trunk was larger than the ROM in the lower trunk after stroke (UT: 12.2°, SD 3.9; LT: 7.9°, SD 2.7; ANOVA, $p<0.001$) [22]. Compared to healthy aged matched individuals, an increase in ROM of the upper trunk of approximately 15% was seen in the stroke group (ANOVA, $F_{1,78}=6.436$, $p<0.02$; Mann-Whitney U-test, $p<0.05$) [22,33]. Lower and whole trunk rotation did not differ between stroke and control (Mann-Whitney U-test, $p>0.05$) according to Wagenaar et al [33]. However, Kerrigan et al [29] found an excessive backward rotation during swing phase in the stroke group compared to healthy individuals (S: 7.6°, SD 8.1; C: 0.8°, SD 1.7°; t-test, $p=0.0006$)

In summary, the majority of studies suggested an increased displacement of the trunk in the frontal and sagittal plane, and larger transversal motion of the upper trunk.

Trunk coordination. Only two studies described the continuous relative phase which occur in the transverse plane (CRP) [22,33]. In subjects suffering from stroke, the transverse plane CRP values were lower (more in-phase) than in the healthy subjects (ANOVA, $F_{1,39}=4.082$, $p<0.05$)[22]. CRP values were higher (more antiphase) in stroke participants with lower gait impairments indicating more dissociation of the thoracic and pelvic segments in good walkers (Spearman correlation, r =

0.63, p<0.05) [22]. Wagenaar et al [33] only mentioned a lack of timing during counter rotation of the trunk in patients suffering from stroke, but found no phase differences between stroke and control (Mann-Whitney U-test, p>0.05).

Trunk variability. Firstly, trunk symmetry which compares trunk displacements or accelerations between the paretic and non-paretic side was examined by four studies [18,20,23,32].

In the frontal plane, lateral trunk accelerations and displacements of the lower trunk showed more asymmetry compared to healthy individuals (independent t-test, t=3.8; p<0.001) [20,32]. Bujanda et al [20] suggested that greater lateral accelerations of the trunk on the paretic side were present. Although no significant differences were found by Dodd et al [23] in lateral displacements of the lower trunk between stroke patients and controls, the stroke group demonstrated a small trend for larger deviations towards the non-paretic foot. The asymmetry index of the mediolateral trunk accelerations was able to discriminate between controls and patients suffering from stroke (AUC=0.76, p=0.001) [32].

In the sagittal plane, the anteroposterior symmetry index showed significant differences between stroke and controls (independent t-test, t=3.91; p<0.001) [32]. Even after controlling for age and gender, differences in asymmetry remained significant (independent t-test, t=-0.06, p<0.01) [32]. Roerdink et al [32] described a difference in magnitude, and not in direction. Patients displaced the trunk farther forward during the paretic step compared to the non-paretic step[18]. The asymmetry index of the anteroposterior trunk accelerations was able to discriminate between controls and patients suffering from stroke (AUC=0.82, p<0.001).

At last, vertical displacements of the trunk were characterized by asymmetry compared to healthy individuals (independent t-test, t=5.06, p<0.001) [32]. The highest discriminating ability was found for the vertical trunk acceleration parameter that classified 85% of the subjects in the hemiplegic group correctly (AUC=0.90, p<0.0001)[32].

Secondly, trunk stability was investigated by five studies of which four concluded that more trunk instability was seen after stroke compared to controls [19,22,30,31]. In both the frontal and sagittal plane, and during vertical accelerations, normalized RMS values of stroke patients were higher than those of the control group, indicating that stroke patients exhibited greater fluctuations of trunk acceleration during walking than the control group [30,31]. Additionally, post-stroke individuals walked with greater local and orbital instability than the controls but remained orbitally stable since orbital instability faded after successive strides (ANOVA, local: p=0.002, orbital: p=0.041) [19]. Only in the transverse plane no differences in the continuous relative phase variability between patients and controls were found (ANOVA, p>0.05) [22].

In summary, the majority of studies suggested a significant increase in trunk variability. Stroke patients had marked asymmetry between paretic and non-paretic sides, and increased instability of the trunk compared to controls.

Effect of walking speed on trunk kinematics. Nine studies examined the effect of walking speed on trunk kinematics [19-24,27,30,33]. First, contradictory information was found regarding linear and angular kinematics.

In the frontal plane, two studies suggested a significant effect of walking speed on lateral displacements [20,23]. Greater lateral displacements were seen when velocity decreased. However, one studies concluded that there was no effect on lateral displacements [21]. In addition, Tyrell et al [24] who investigated the effect of walking speed on pelvic obliquity did not found any differences between various speed conditions.

No data regarding the sagittal plane was collected.

In the transversal plane, the majority of studies concluded that rotations in the lower trunk were affected by walking speed ($F_{1.78}$=4.168, ANOVA, $p<0.05$) [22,27]. Although Hacmon et al [22] stated that trunk rotation seemed to decrease at a higher speed, Ford et al [27] concluded that trunk rotation significantly increased with walking speed. Additionally, Wagenaar et al [33] did not find any significant differences.

Second, research is too scarce concerning the continuous relative phase difference between the upper and lower trunk to form conclusions. Although both studies suggest that a more in-phase rotation is seen when walking at a slower speed, only one study found significant differences [22,33].

Third, movements of trunk seemed to be more stable and symmetrical if walking velocity increased in all three planes (Pearson's correlation, F: $r=0.646$, $p=0.09$; S: $r=0.687$, $p=0.005$; V: $r=0.838$, $p<0.001$) [19,30]. The RMS measuring vertical accelerations seemed to be the parameter most dependent on walking velocity in all three groups [30]. Only one study found no effect of walking speed on trunk symmetry ($r=-0.14$, ND, $p=0.13$) [23].

In summary, although the majority studies suggested that a more normal walking pattern was possible when walking speed was increased, not every study found significant results or agreed on the magnitude or direction of movement.

Discussion

Summary of evidence. The purpose of this study was to gain more insight in trunk biomechanics in stroke patients compared to healthy individuals and to investigate the effect of walking speed on trunk biomechanics.

The overall risk of bias of the included studies was moderate, both the cross-sectional as the case-control studies had a median score of six. In the case-control studies, none of the studies were blinded to the case or control status. Since gait analysis is a highly objective method to quantify locomotion, it is of less importance in assessing risk of bias. However, the majority of studies were not able to control for confounding variables or had a sample size which was unsatisfactory and unjustified.

Results of this review indicate that kinematic parameters are altered during hemiplegic gait and show increased trunk motion in the lateral and sagittal plane, larger upper trunk motion in the transverse plane, decreased antiphase rotation of the upper and lower trunk, and less stability and symmetry compared to non-pathological gait. Insufficient data were found concerning muscle activity and kinetics. Similar results were found in other patient populations. For example, children with cerebral palsy also showed increased ROM and decreased antiphase rotation during walking [4,5]. Moreover, patients with Parkinson's disease and low back pain had a decrease in antiphase trunk rotation [34,35].

These alterations in trunk movement cannot solely be attributed to an impairment in trunk function. Indeed, trunk muscles are impaired after stroke by reduced activity levels, delayed onset of and reduced synchronization [8]. Yet, many of these altered trunk movements can also be considered compensatory for lower limb impairments and other gait problems. These other problems leading to altered trunk movements can be diverse in stroke patients. Lack of foot clearance results from insufficient hip flexion, knee flexion, and dorsiflexion and is associated with an upward pelvic movement (hip hiking) to ensure sufficient clearance of the affected foot [36,37]. This can be the result of muscle weakness or a synergistic extension motor pattern in the lower limbs. In addition, excessive lateral trunk motion can be caused by weak abductor muscles [38]. The so-called Trendelenburg sign is positive when a pelvic drop is apparent at the swing leg during single stance by weakness of the gluteus medius muscle of the stance leg. A patient is able to compensate for this drop by leaning over to the affected side. Since several lower limb impairment can cause trunk deviations, it is of great importance to differentiate between compensatory trunk movements and intrinsic trunk control deficits. Several of the included studies examined the association of motor recovery of the lower limbs with kinematic impairments of the trunk. A moderate to strong negative correlation was found between motor recovery of the leg and foot with transverse ROM of the trunk

[21,22], and between paretic hip flexion, hip extension, knee flexion and frontal pelvic motion [28]. Moreover, a strong positive correlation between hip adduction and knee extension torques, and frontal pelvic motion [28]. However, the level of spasticity in the lower limbs seemed to be less correlated with trunk impairments [21]. This might suggest that various trunk deviations in ROM in the frontal and transverse plane might be compensatory. Further research is necessary to establish a causal link and to further examine associations with trunk coordination and trunk variability, since none of the studies examined this. However, clinicians are still advised to look at the individual level of recovery of the patient before creating a specific treatment plan.

Alterations in trunk motion should be treated according to the underlying cause. Intrinsic trunk control deficits should be recognised and treated by a specific trunk control exercise program, since truncal exercises are able to improve trunk performance and balance [39]. On the other hand, compensatory trunk movements may not require specific trunk rehabilitation and can be resolved by treating the already mentioned underlying deficits. Shoes, orthoses and assistive devices might be able to treat the underlying deficits by altering lower limb kinematics, spatiotemporal parameters, and stability [40-42]. First of all the type of shoe can alter spatiotemporal parameters. Close fitting shoes resulted in a higher walking speed and increased step length compared to slippers or barefoot [41]. In the current review, the majority of studies reported that patients could wear their own comfortable shoes, which might have influenced the results. However, no studies have been conducted concerning the effect of the type of shoe on trunk biomechanics. Second, ankle foot orthoses (AFO) are able to facilitate weight-bearing on the hemiplegic leg, increase stability during stance, and reduce energy cost of walking [40]. More important, AFO's may provide sufficient clearance during walking when a foot drop is apparent. As a result, compensatory movements such as increased pelvic tilt or hip hiking are unnecessary to provide clearance and are therefore able to decrease the alterations seen during hemiplegic gait. Third, the use of a cane can increase pelvic obliquity during the stance phase in the hemiplegic limb and stability during single limb support [42]. In the current review, not all studies allowed assistive devices and orthoses during gait analysis. Therefore, some studies might have eliminated the compensatory trunk movements by allowing assistive devices and/or orthoses compared to others who examined both compensatory movements and intrinsic trunk deficits by not allowing these devices. However, since not all studies specified the walking conditions in such a manner it was hard to exclude studies based on the use of orthoses and/or assistive devices during the dynamic walking trials.

Another important finding of this study is the influence of walking speed on trunk kinematics as total trunk motion and trunk coordination seemed to be speed-dependent. The majority of studies concluded that a more normal biomechanical trunk pattern was observed when walking at higher speeds. Our results can be elucidated by the findings of Hak et al [43], they suggested that a lower walking speed in post-stroke individuals may cause a decrease in the backward margins of stability and an increase in the risk of falls. Even in healthy individuals and patients with Parkinson's disease alterations in trunk movement are seen when walking speed is altered [13,35]. At slower speed, stroke patients tend to move their trunk into a more in-phase coordination compared to walking at higher speed. In contrary, an increase in antiphase rotation induced by a difference in pelvic timing was observed in healthy individuals [12,13]. The pelvis seemed to move more in-phase with the lower limbs when walking at a higher speed. So, it seems that stroke patients adopt another strategy than healthy individuals. However, this is not entirely true. It is important to consider that the maximum walking speed for stroke patients in the included studies was approximately 1 m/s compared to 1.5 m/s for healthy adults. Consequently, it might be that patients are not able to reach the certain walking speed which induces the pelvic step. As already mentioned, at a certain walking speed pelvic rotations start to contribute to step length which was defined in literature as the pelvic step. People suffering from stroke show a similar walking pattern at maximal speed as healthy individuals when walking at their self-selected speed. Therefore, increasing walking speed normalizes the gait pattern of stroke patients. However, is seems not possible to increase walking speed in such a fashion that the pelvic step is induced.

There are limitations of this review that should be acknowledged. At first, comparison of the included studies was done without considering the different effects of treadmill and over ground walking. Stroke subjects walk over ground with significant higher maximal speeds, greater stride lengths, and lower cadence compared to treadmill-walking [44]. A more detailed comparison was made by Riley et al [45], they suggested that pelvic tilt, pelvic obliquity, pelvic rotation, spine flexion, and spine lateral flexion was significantly different during over ground walking than treadmill walking in healthy subjects. Therefore, comparing different protocols should be done with caution. However, to be able to give an overall picture of the trunk during walking, we decided to include both protocols. Second, recording equipment used in gait analysis differed in the included studies. Some were based on image processing, while others were based on sensors. Sensors allow analysis during activities of daily living and in settings outside the laboratory compared to several image processing techniques. However, 2D motion capture and accelerometers are as reliable and valid as the gold standard of 3D-motion capture. We therefore decided to compare the outcomes generated by the different recording systems. At last, both chronic and sub-acute stroke patients were included in this study. It might be that trunk alterations are time-dependent. For future research, it might be interesting to see if trunk biomechanics alter when time post stroke increases.

Conclusions. The findings from this systematic review suggests that patients after stroke walk with increased mediolateral-anteroposterior trunk movements, more in-phase coordination, and increased instability and asymmetry. However, more research is necessary since a lot of contradictions are still present concerning trunk biomechanics. In addition, walking speed has a considerable effect on these impairments. At last, studies examining trunk kinetics and trunk muscle activity during walking in stroke patients are lacking which necessitates further study. During rehabilitation it is important to correctly identify the cause of the alterations in the trunk. Are these compensatory for lower limb impairments or are these intrinsic trunk deficits? Compensatory movements might be resolved by using orthoses and assistive devices or by threating the underlying cause. However, trunk alterations cannot be exclusively contributed to lower limb impairments. Therefore, it is important to incorporate a trunk training exercises protocol in conventional rehabilitation to influence the intrinsic trunk deficits. At last, several studies suggested that walking beyond the patient's self-selected walking speed induced a more natural pattern in trunk biomechanics. However, further research is necessary to examine the effect of gait rehabilitation beyond the patient's comfortable walking speed since the positive effect on trunk biomechanics might be cancelled out by possible adverse effects on lower limb kinematics or postural control.

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