The influence of a thoracolumbosacral orthosis on gait performance in healthy adults during walking

TAMAYA VAN CRIEKINGE[1, 2]*, STEVEN TRUIJEN[1, 2], ANN HALLEMANS[1, 2], WIM SAEYS[1, 2, 3]

[1] Department of Rehabilitation Sciences and Physiotherapy, Faculty of Medicine and Health Science,
University of Antwerp, Belgium.
[2] Multidisciplinary Motor Centre Antwerp, University of Antwerp, Belgium.
[3] RevArte Rehabilitation Hospital, Edegem, Antwerp, Belgium.

Purpose: Since the thorax and pelvis are primary determinants of normal and pathological walking, it is important to know how gait performance is influenced when the trunk is constraint. The objective of this study is to investigate the effect of a thoracolumbosacral orthosis (TLSO) on gait performance in healthy adults during overground walking. *Methods*: Fourteen healthy volunteers walked with and without TLSO. Outcome measures consisted of spatiotemporal parameters and clinically important joint angular time profiles of the lower limbs. Joint angular time profiles were assessed in the sagittal, frontal and transversal plane. A paired *t*-test was used for discrete parameters and spm1d for assessing the joint angular time profiles. *Results*: Walking with a constraint resulted in decreased stride time and step time, increased step width and cadence. In the sagittal plane, no significant differences were observed regarding joint kinematics in the hip, knee and ankle. In the frontal plane, decreased adduction during stance and abduction during swing was observed in the hip. In the transversal plane, increased external rotation of the hip and increased internal rotation of the ankle was seen when wearing a contstraint. *Conclusions*: Wearing a TLSO can already bring forth significant changes in gait performance, suggesting an important relationship between trunk movements and mobility.

Key words: torso, orthotic devices, locomotion, kinematics, biomechanical phenomenon

1. Introduction

The pelvis is thought to be of high importance during locomotion, as Saunders, Inman and Eberhart [1] described six determinants of normal gait, of which three were related to pelvic movement [1], [2]. The pelvis is considered to be a major part of the trunk, together with the thorax. Although thorax and pelvis move in a coordinated manner around the same vertical axis, the thorax rotates in the opposite direction as the pelvis during normal gait. This movement pattern is also known as anti-phase rotation which controls total body angular momentum and improves movement efficiency [3], [4]. A total net angular momentum close to zero was shown during walking, providing evidence that the motion of the thorax cancels out the movement of the pelvis [4].

Changes in trunk motion, such as increased trunk stiffness and in-phase coordination of the pelvis and thorax during walking are associated with aging and many movement disorders [5]–[7]. In-phase coordination is characterized by a phase difference of zero, thus by little to no dissociation of the pelvis and thorax. Research regarding the influence of a thoracolumbo-

sacral orthosis (TLSO), which impairs thorax-pelvic dissociation kinematics and increases trunk stiffness, on gait performance during overground walking is scarce. A TLSO is able to artificially induce those age-related changes in trunk performance. The results of this study could lead to a better understanding of the importance of trunk movements during aging. Since aging is associated with several other processes such as sarcopenia, cognitive decline, sensory changes and even motor coordination, age-related changes in gait and its relation to trunk motion is still unclear. Experimental designs in which the pelvis is fixated by an orthosis or brace resulted in a significantly reduced stride and step length, walking speed, percentage stance phase and step width variability, and increased step width [8]–[10]. Although these studies demonstrate short-term gait changes during walking, none of the previous studies investigated the effect of a total trunk constraint, fixation of thorax and pelvis during walking. Using a total trunk constraint, increased external work and changes in thorax/pelvis-leg relative phase have been described [11]–[13]. However, in these studies, participants walked on a treadmill, which induces differences in temporal and angular kinematics, compared to overground walking [14]. Only three studies examined the effect of artificial stiffening of the trunk during overground walking [15]–[17]. When wearing the trunk constraint, individuals walked with shorter steps, and had a reduced single support phase [15]. However, the device used in the study of Russel, Kelleran and Morrison [15] was a Kendrick Extrication Device, which is used for immobilization of the head, neck and torso, not the pelvis. Moreover, in the study of Konz, Fatone and Gard [16], and Song, Kim and Kim [17] only young and healthy individuals were included, mean age was 27 and 25 years, respectively. Since increased trunk stiffness and in-phase coordination are typically seen in older adults, it is also important to include subjects over the age of 30 years [5]–[7]. Therefore, more research is necessary to examine the effect of a TLSO on gait performance during overground walking in healthy adults with a broader age range.

Gait performance can be assessed by several gait characteristics, such as spatiotemporal parameters (STP). A combination of shorter step length and increased step width are commonly described as a strategy to cope with imbalance [18]. This imbalance might result in a lack of trunk control, since variations in trunk sway are sensitive enough to detect changes in postural control [19]. Therefore, constraining the trunk does not only prevent typical anti-phase coordination, but also alters balance control strategies. There are three typically used balance strategies, ankle, hip and stepping strategy. Due to the constraint, movements of the hip are limited, which alters the possibility to react on balance disturbances. On the other hand, the constraint also provides more stability which could, therefore, increase balance. Changes of increased or decreased balance might cause compensatory reactions in STP to counteract the changes in postural control, which can tell us more about the role of the trunk in postural balance control.

The aim of this study was to investigate the effects of a TLSO on gait performance in healthy adults during overground walking. We hypothesized that several compensatory mechanisms will arise to maintain an efficient and balanced walking pattern. Concerning STP, we speculate that decreased step length, increased step width and step variability, and a lower walking speed will be the result of trunk constraint walking since these compensations are related to imbalance. Moreover, to understand STP differences, kinematic parameters should also be investigated. Therefore, clinically important kinematic parameters of the hip, knee and ankle, were investigated in the sagittal, frontal and transverse plane.

2. Materials and methods

This review was conducted according to the STrengthening the Reporting of OBservational studies in Epidemiology (STROBE) statement.

Setting

Participants received an instrumented gait analysis performed at a movement analysis laboratory equipped with an automatic three-dimensional motion capture (Vicon T10, sampling rate 100 Hz., ©Vicon Motion Systems Ltd., Oxford, UK, 100 frames per second, resolution 1 Megapixel 1120 × 896), 3 AMTI type OR 6–7 force plates (1000 frames per second, 46 × 50 × 8 cm, ©Advanced Mechanical Technology, Inc., Watertown, USA) and 1 AccuGait® force plate (1000 frames per second, 50 × 50 × 4 cm, ©Advanced Mechanical Technology, Inc., Watertown USA). Reflective markers were attached to anatomical landmarks on the participant's body according to the standard Plug-In-Gait model [20]. Participants subsequently walked barefoot over a 12-meter walkway at a self-selected speed with and without trunk constraint (Fig. 1). In total, a mini-

mum of six walking trials were recorded per condition. The study protocol was approved by the local ethics committee according to the Declaration of Helsinki. Data collection lasted from September 2013 until June 2014.

Participants

In total, 14 healthy adults ranging from 20 to 60 years old enrolled in the study. Volunteers were excluded from the sample if they had self-reported visual impairments, antalgic gait pattern, abnormal mobility in the lower limbs or any known neurological or orthopaedic disorder that could influence motor performance and balance. Informed consent was obtained from all subjects prior to participation.

Variables of interest

Demographic data (sex, age, body length, body mass and leg length) were used to describe the sample. The mean and standard deviations (SD, variability) of step and stride time [s], -length [m], -width [m] and stance [%], and mean walking speed [m/s] and cadence [steps/min] were selected as outcome variables. Those variables were calculated as the mean of two trials in each condition. Per trial, the mean of at least three strides within one trial was used. Step variability measures were determined from inter-individual variations, expressed as SD, between strides and trials. Step length was normalized by dividing the subject's step length by their leg length, thus controlling for differences in limb length. In addition, clinically important joint angular time profiles of the hip, knee, ankle and foot in the sagittal, frontal and transversal plane were observed. At last, to make sure that the constraint fixated the trunk properly, range of motion (RoM) of the thorax and pelvis was calculated in all three planes. The difference between the maximum and minimum angles during stance and swing phase was defined as RoM.

Measurements and data calculations

For each subject, body mass, height, leg length (right leg) and joint width were collected according to standard procedures of the Plug-In-Gait model [20]. Markers were attached at anatomical landmarks according to the same procedures [20]. However, when participants were asked to wear the trunk constraint, sternum and both the left and right spina iliaca anterior superior were no longer visible. Hence, measurements of these anatomical landmarks were carefully taken in order to attach the markers on the trunk constraint at the precise location. To minimize movement

of the brace and thus of reflective markers, four different sizes of constraints were available to assure a perfect fit (small, medium, large and extra-large). Reflective markers were tracked and labelled using the Vicon Nexus 1.8.5 software. Trajectories were filtered (low pass zero phase shift 4th order Butterworth filter, cut-off frequency 6 Hz.) and the Vicon Plug-In Gait model[®] was used. Based on the heel marker trajectories and force plate recordings, events of foot strike and foot off were determined. Once all markers were visible for at least three consecutive strides, trials were further processed. The .c3d files obtained in Vicon Nexus 1.8.5 were exported to a custom made MATLAB[®] (R2015a for Windows, ©The MathWorks, Inc., Natick, USA) model to calculate the variables of interest. STP were calculated based on the left and right ankle marker trajectories and generated with the "Gait Cycle Parameters Pipeline Operation" of the Vicon Nexus software.

Fig. 1. Trunk constraint

Statistical analysis

Statistical analysis was performed using SPSS version 24[®] for Windows (©IBM Corporations, New York, USA). Descriptive statistics were performed to characterize the study sample and the Shapiro–Wilk test was performed to test for normality of distribution. A student t-test was used to compare left and right STP, differences between the two conditions, and to confirm whether the trunk constraint was fixated properly. At last, joint angular time profiles were assessed across the entire gait cycle using spm1d, paired t-test. This technique not only detects changes in continuous data sets, but also conducts statistical hypothesis testing in a continuous manner, directly on the original curves. Spm1d has been thoroughly described by Pataky et al. [21], [22]. Significance level was set at $p < 0.05$.

3. Results

Descriptive characteristics

Eight healthy men and six healthy female volunteers were included in this study (Table 1). Mean age was 39 years (SD 14 years), six individuals were aged between 20 and 30 years, four subjects were aged between 40 and 50 years, and four were over the age of 50 years. Since no significant differences were found between left and right STP in the no constraint condition ($P > 0.05$), only parameters of the right stride were further analysed.

Table 1. Characteristics of the participants

	N	Range	Mean	SD
Body Height [mm]	14	1605–1855	1754.64	75.25
Body Mass [kg]	14	57.30–94.50	76.91	12.05
Leg Length [m]	14	0.83–0.97	0.96	0.04
Age [years]	14	21–57	39.35	14.23
Gender	14	6F/8M		

F – female, M – male, SD – standard deviations.

Trunk motion

RoM of the thorax significantly differed in the sagittal plane. Anteroposterior movements were decreased when wearing a trunk restraint ($P = 0.014$). RoM of the pelvis significantly decreased In all three planes when wearing the constraint ($P < 0.05$). This suggests that the TLSO was successful in fixating the trunk.

Spatiotemporal parameters

Significant differences between the two conditions were found. Wearing the TLSO resulted in decreased stride ($P = 0.04$) and step time ($P = 0.013$), increased step width ($P = 0.002$) and higher cadence ($P = 0.004$) (Table 2). No significant differences were observed in the variability measures.

Joint Angular Time Profiles

Joint angular time profiles of the hip, knee, ankle and foot are shown in Fig. 2. In the sagittal plane, no significant differences were observed regarding joint kinematics in the hip, knee and ankle over the entire gait cycle. In the frontal plane, significant differences in the hip were observed after contralateral (10–20% of gait cycle) and ipsilateral (60–70% of gait cycle) toe off, when comparing walking with and without TLSO. Peak amplitudes decreased significantly after contralateral ($t = 3.007$, $P = 0.021$) and ipsilateral toe off ($t = 3.007$, $P = 0.018$), suggesting that there is decreased adduction during stance and abduction during swing. In the transverse plane, significant differences in the hip were observed during mid and terminal stance (20–40% of gait cycle) and during initial and mid swing (65–87% of gait cycle), when comparing the two walking conditions. Walking with TLSO resulted in increased external rotation of the hip over the entire gait cycle. However, differences were only significant during mid-terminal stance ($t = 3.194$, $P = 0.001$) and initial-mid swing ($t = 3.194$, $P < 0.001$). Even though the foot progression ankle did not differ between the two conditions, significant differ-

Table 2. Comparison spatiotemporal parameters with and without constraint

	Without constraint (mean, SD)	With constraint (mean, SD)	*p*-value
Mean stride time [s]	1.096 ± 0.746	1.051 ± 0.819	0.004**
Mean step time [s]	0.555 ± 0.350	0.526 ± 0.418	0.013*
Mean normalized stride length [m]	1.459 ± 0.138	1.432 ± 0.103	0.249
Mean normalized step length [m]	0.727 ± 0.697	0.718 ± 0.612	0.484
Mean step width [m]	0.172 ± 0.176	0.186 ± 0.217	0.002**
Mean stance [%]	61.064 ± 1.653	61.182 ± 1.463	0.484
Mean gait speed [m/s]	1.219 ± 0.139	1.237 ± 0.133	0.550
Mean cadence [steps/min]	107.905 ± 7.762	113.262 ± 9.423	0.004**
Variability stride time [s]	0.023 ± 0.173	0.021 ± 0.127	0.743
Variability step time [s]	0.024 ± 0.134	0.018 ± 0.089	0.156
Variability normalized stride length [m]	0.029 ± 0.295	0.036 ± 0.287	0.584
Variability normalized step length [m]	0.020 ± 0.111	0.021 ± 0.121	0.869
Variability in step width [m]	0.026 ± 0.140	0.026 ± 0.145	1.000
Variability stance [%]	1.644 ± 1.144	1.286 ± 0.876	0.357

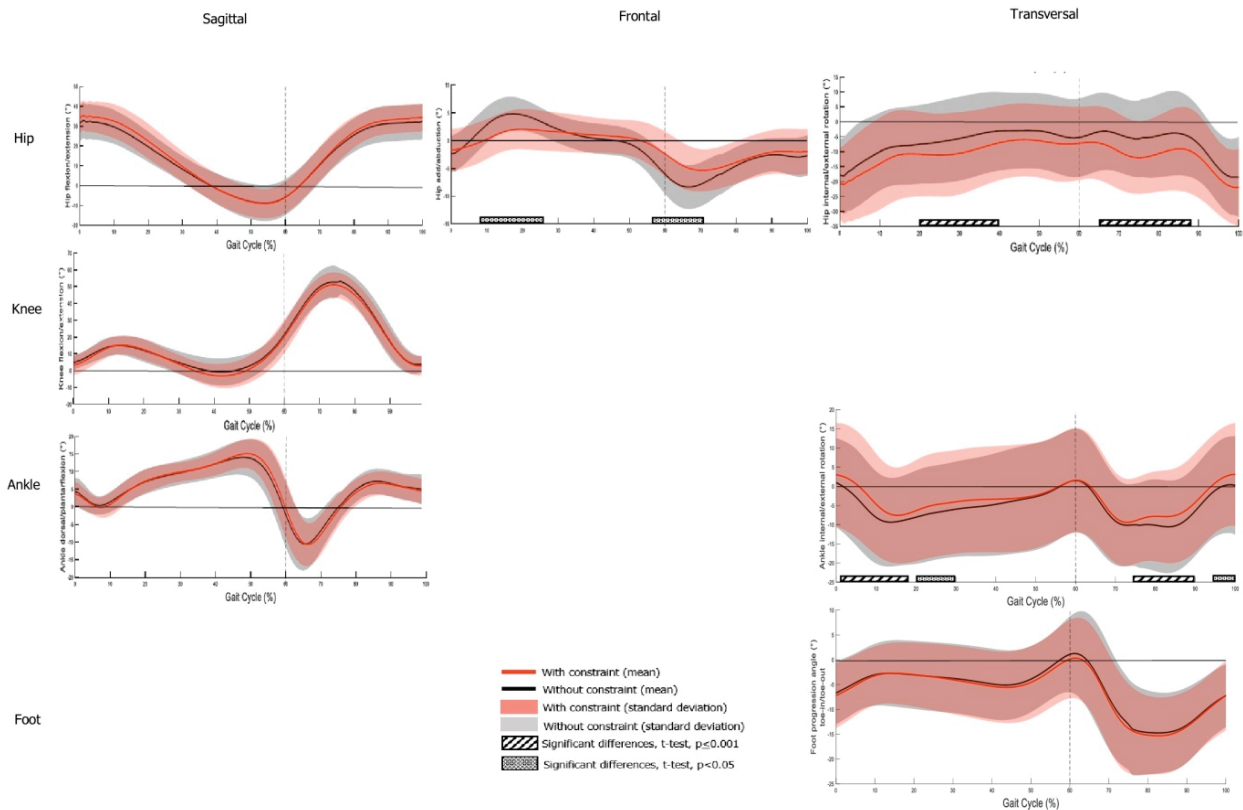SD – standard deviation, *$p < 0.05$, ** $p < 0.01$.

Fig. 2. Joint Angular Time Profiles

ences were found in ankle rotation during loading response and mid stance (3–30% of the gait cycle), mid-terminal swing (75–90% and 95–100% of gait cycle). Walking with TLSO resulted in increased internal rotation of the ankle during loading response ($t = 3.384$, $P < 0.0001$), mid stance ($t = 3.384$, $P < 0.012$), mid swing ($t = 3.384$, $P = 0.001$), and terminal swing ($t = 3.384$, $P = 0.037$).

4. Discussion

Summary of evidence

Since thorax and pelvis are the primary determinants of normal and pathological walking, it is important to know how gait performance is influenced when age-related changes at the level of the trunk occur. Since the utilised trunk constraint was effective in fixating the thorax and pelvis, healthy individuals were able to walk with artificially-induced trunk stiffness. Therefore, we were able to investigate the effect of a TLSO on gait performance in healthy adults.

The results of this study showed that the TLSO induced differences in healthy adults, stride and step time decreased by approximately 5 and 4%, and mean step width and cadence increased by 8% and 5%, re-

spectively. These changes are solely due to stiffening of the trunk, yet aging coincides with several other processes (e.g., cognitive decline, muscle weakness, etc.) influencing gait performance. Therefore, 4 to 8% of the changes seen in gait performance of the elderly are caused by the age-related changes in trunk movements. In addition, if there are already significant differences in a healthy population, one must consider the fact that when the trunk is severely impaired, e.g., in stroke and Parkinson's disease, greater changes will be observed in gait performance. For example, patients with both Parkinson's disease and low back pain show changes in trunk coordination and decreased antiphase rotation in the trunk [23], [24]. Moreover, patients suffering from stroke walk with decreased antiphase rotation and enhanced trunk motion in the lateral and sagittal plane [25]. In this study, induced changes in trunk motion, when comparing walking with or without constraint, ranged from 2 to 66%. On the other hand, changes over 66% are seen in trunk motion after a stroke, compared to healthy individuals [25]. One can only assume that greater impairments in the trunk could lead to clinically important changes in gait performance. However, as the trunk impairments seen in these patient population do not solely result from trunk stiffening, impaired neuromuscular control will be an important predictor which also has

a direct effect on gait performance. Further research is necessary to examine the influence of trunk impairments on gait performance in these patient populations.

Counter rotations of the pelvis and trunk during normal walking result in the control of total body angular momentum and improved movement efficiency [3], [4]. As a result, walking with a trunk constraint might result in a decreased movement efficiency and a greater total body angular momentum [26]. This momentum is created when trunk and pelvic segment move, more or less, in the same direction [26]. The increased total body angular momentum might result in a faster lower limb propulsion, and therefore decreasing step and stride time. Of course, by reducing step and stride time, subjects are able to take more steps over a fixed period of time, which results in a higher cadence. Under other conditions, the decline in movement efficiency during constraint walking expressed itself as a decrease in postural control which was compensated by increased step width. The first prerequisite of gait described by Perry [27] is "stability during stance", which is the ability to maintain the centre of mass inside the base of support and provide sufficient balance. People are constantly altering the position of the trunk to stay balanced over the base of support. The inability to correct the trunk's position might lead to a compensatory strategy in which an enlarged base of support is created. This compensatory strategy is characterised by increasing step width which might ensure that the centre of mass is located inside the base of support. Nevertheless, it was quite surprising that no differences were observed in stride or step length since there is a strong relation between pelvic rotations and step length. Literature has introduced the concept "pelvic step" which describes that horizontal pelvic rotations contribute to step length from a certain walking speed [28]. A more recent study concluded that the contribution of pelvic rotations to step length are rather small, maximally 3% [29]. Liang et al. [29] suggest that the pelvic step only contributes to step length when walking more in-phase and with a walking speed above 0.83 m/s, both requirements are met in our study. So, we expected that the concept of the pelvic step would apply here. Two reasons can explain why no differences were observed: 1) The limited amount of fixation of the pelvis, a decrease in transverse RoM was only 29%, which might be too small to induce differences in step length; or 2) Step length is initially created by hip extension, since no changes in RoM were seen in this joint, there was no need for a compensatory strategy as "pelvic step".

Another important finding is that significant differences were seen in frontal and transversal kinematics of the hip. Wearing the constraint resulted in decreased frontal range of motion and increased external rotation of the hips. Increased step width mostly coincided with decreased adduction of the hips during stance. Participants walked with more external rotation, however, the foot progression angle did not differ in the two conditions. The lack of toeing-out was compensated by increased internal rotation of the ankle when wearing the constraint. This suggests that the trunk is not a passenger which is carried by the lower limbs, but an important segment actively contributing to gait. Impairments solely located at the level of trunk can already bring forth significant changes in gait performance.

Although this study concludes that there is an important relationship between trunk and gait performance, further research is necessary to examine the type, direction and magnitude of this relationship. By doing this, selecting specific gait parameters might predict the involvement of trunk impairments or vice versa. However, to fully understand this relationship, it is important to investigate the influence of trunk impairments on gait performance in several patient populations facing these kind of impairments.

There are some important limitations to consider. First, the sample size may have been too small, and further larger studies are required to confirm these results. Second, since smp1d is a fairly new statistical tool to examine a continuous dataset, analyses are still under development and should be interpreted with caution. Yet, resulting errors are expected to be small [24], [25]. Third, when the trunk constraint was worn, markers were placed onto the constraint. Although, several constraints were present to ensure a good fit, a small standardized measurement error might be present in all individuals at the level of the trunk.

5. Conclusions

The use of TLSO in healthy adults resulted in a decreased stride and step time, and increased step width and cadence, suggesting an important relationship between trunk and gait performance. Age-related changes, such as increased trunk stiffness which are solely located at the level of trunk, can already bring forth significant changes in gait performance. To fully understand this relationship, future research should investigate how actual trunk impairments influence gait performance and if some gait deviations are distinctive for specific trunk deficits in specific patient populations.

Clinical relevance

Wearing a TLSO changed locomotion in healthy adults, suggesting the importance of trunk movements during walking. Trunk movements are responsible for 4–8% of the variations in spatiotemporal parameters and for kinematic changes in the frontal and transversal plane. Therefore, trunk motion should be incorporated in traditional gait assessment.

References

[1] SAUNDERS J.B., INMAN V.T., EBERHART H.D., *The major determinants in normal and pathological gait*, J. Bone Joint Surg. Am., 1953, 35-A(3), 543–558.

[2] LEWIS C.L., LAUDINCIA N.M., KHUU A., LOVERRO K.L., *The human pelvis: variation in structure and function during gait*, Anat. Rec. (Hoboken), 2017, 300(4), 633–642.

[3] HERR H., POPOVIC M., *Angular momentum in human walking*, J. Exp. Biol., 2008, 211(Pt 4), 467–481.

[4] SCHONER G., JIANG W.Y., KELSO J.A., *A synergetic theory of quadrupedal gaits and gait transitions*, J. Theor. Biol., 1990, 142(3), 359–391.

[5] SWINNEN E., BAEYENS J.P., PINTENS S., BUYL R., GOOSSENS M., MEEUSEN R. et al., *Walking more slowly than with normal velocity: The influence on trunk and pelvis kinematics in young and older healthy persons*, Clin. Biomech., 2013, 28(7), 800–806.

[6] ADKIN A.L., BLOEM B.R., ALLUM J.H., *Trunk sway measurements during stance and gait tasks in Parkinson's disease*, Gait Posture. 2005, 22(3), 240–249.

[7] MIZUIKE C., OHGI S., MORITA S., *Analysis of stroke patient walking dynamics using a tri-axial accelerometer*, Gait Posture, 2009, 30(1), 60–64.

[8] VENEMAN J.F., MENGER J., VAN ASSELDONK E.H., VAN DER HELM F.C., VAN DER KOOIJ H., *Fixating the pelvis in the horizontal plane affects gait characteristics*, Gait Posture, 2008, 28(1), 157–163.

[9] DONELAN J.M., SHIPMAN D.W., KRAM R., KUO A.D., *Mechanical and metabolic requirements for active lateral stabilization in human walking*, J. Biomech., 2004, 37(6), 827–835.

[10] MUN K.R., GUO Z., YU H., *Restriction of pelvic lateral and rotational motions alters lower limb kinematics and muscle activation pattern during over-ground walking*, Med. Biol. Eng. Comput., 2016, 54(11), 1621–1629.

[11] MAHAUDENS P., BANSE X., DETREMBLEUR C., *Effects of short-term brace wearing on the pendulum-like mechanism of walking in healthy subjects*, Gait Posture, 2008, 28(4), 703–707.

[12] WU W.H., LIN X.C., MEIJER O.G., GAO J.T., HU H., PRINS M.R. et al., *Effects of experimentally increased trunk stiffness on thorax and pelvis rotations during walking*, Hum. Mov. Sci., 2014, 33, 194–202.

[13] TAIAR R., MITTON A., CAMBIER J., GUENAUOUI I., MICHEL S., ABDI E., POLIDORI G., *Biomechanics of the immediate impact of wearing a rigid thoracolumar corset on gait kinetmatics and spatiotemporal parameters*, MATEC Web of Confernces, 2018, 145, 04007.

[14] CHOCKALINGAM N., CHATTERLEY F., HEALY A.C., GREENHALGH A., BRANTHWAITE H.R., *Comparison of pelvic complex kinematics during treadmill and overground walking*, Arch. Phys. Med. Rehabil., 2012, 93(12), 2302–2308.

[15] RUSSEL D.M., KELLERAN K.J., MORRISON S., *Bracing the trunk and neck in young adults leads to a more aged-like gait*, Gait Posture, 2016 Sep., 49, 388–393.

[16] KONZ R., FATONE S., GARD., *Effect of restricted spinal motion on gait*, J. Rehabil. Res. Dev., 2009, 43(2), 161–170.

[17] SONG H.N., KIM Y.M., KIM K., *A kinematic analysis of the lower limb with regard to restricted spinal motion during gait*, J. Phys. Ther. Sci., 2017, 29(1), 81–84.

[18] HAK L., HOUDIJK H., STEENBRINK F., MERT A., VAN DER WURFF P., BEEK P.J., VAN DIEËN J.H., *Speeding up or slowing down?: Gait adaptations to preserve gait stability in response to balance perturbations*, Gait Posture, 2012, 36(2), 260–264.

[19] GILL J., ALLUM J.H., CARPENTER M.G., HELD-ZIOLKOWSKA M., ADKIN A.L., HONEGGER F., PIERCHALA K., *Trunk sway measures of postural stability during clinical balance tests: effects of age*, J. Gerontol. A Biol. Sci. Med. Sci., 2001, 56(7), M438–M447.

[20] DAVIS R.B., OUNPUU S., TYBURSKI D., GAGE J.R., *A Gait Analysis Data-Collection and Reduction Technique*, Hum. Movement Sci., 1991, 10(5), 575–587.

[21] PATAKY T.C., ROBINSON M.A., VANRENTERGHEM J., *Region-of-interest analyses of one-dimensional biomechanical trajectories: bridging 0D and 1D theory, augmenting statistical power*, Peer J., 2016, 4, e2652.

[22] PATAKY T.C., VANRENTERGHEM J., ROBINSON M.A., *The probability of false positives in zero-dimensional analyses of one-dimensional kinematic, force and EMG trajectories*, J. Biomech., 2016, 49(9), 1468–1476.

[23] VAN EMMERIK R.E., WAGENAAR R.C., WINOGRODZKA A., WOLTERS E.C., *Identification of axial rigidity during locomotion in Parkinson disease*, Arch. Phys. Med. Rehabil., 1999, 80(2), 186–191.

[24] SEAY J.F., VAN EMMERIK R.E., HAMILL J., *Low back pain status affects pelvis-trunk coordination and variability during walking and running*, Clin. Biomech., 2011, 26(6), 572–578.

[25] VAN CRIEKINGE T., SAEYS W., HALLEMANS A., VELGHE S., VISKENS P., VEREECK L. et al., *Trunk biomechanics during hemiplegic gait after stroke: A systematic review*, Gait Posture, 2017(54), 133–143.

[26] PREECE S.J., MASON D., BRAMAH C., *The coordinated movement of the spine and pelvis during running*, Hum. Mov. Sci., 2016, 45, 110–118.

[27] PERRY J., *Normal and pathological gait*, Atlas of Orthotics, 2nd ed., St. Louis: C.V. Mosby, pp. 76–111.

[28] HUANG Y., MEIJER O.G., LIN J., BRUIJN S.M., WU W., LIN X. et al., *The effects of stride length and stride frequency on trunk coordination in human walking*, Gait Posture, 2010, 31(4), 444–449.

[29] LIANG B.W., WU W.H., MEIJER O.G., LIN J.H., LV G.R., LIN X.C. et al., *Pelvic step: the contribution of horizontal pelvis rotation to step length in young healthy adults walking on a treadmill*, Gait Posture, 2014, 39(1), 105–10.