Journal of Biomechanics 47 (2014) 2263–2268

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Contributions of individual muscles to the sagittal- and frontal-plane angular accelerations of the trunk in walking

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ARTICLE INFO

Article history:
Accepted 26 April 2014

Keywords:
Trunk
Walking
Induced acceleration
Dynamic simulation
Muscle contributions

ABSTRACT

This study was conducted to analyze the unimpaired control of the trunk during walking. Studying the unimpaired control of the trunk reveals characteristics of good control. These characteristics can be pursued in the rehabilitation of impaired control. Impaired control of the trunk during walking is associated with aging and many movement disorders. This is a concern as it is considered to increase fall risk. Muscles that contribute to the trunk control in normal walking may also contribute to it under perturbation circumstances, attempting to prevent an impending fall. Knowledge of such muscles can be used to rehabilitate impaired control of the trunk. Here, angular accelerations of the trunk induced by individual muscles, in the sagittal and frontal planes, were calculated using 3D muscle-driven simulations of seven young healthy subjects walking at free speed. Analysis of the simulations demonstrated that the abdominal and back muscles displayed large contributions throughout the gait cycle both in the sagittal and frontal planes. Proximal lower-limb muscles contributed more than distal muscles in the sagittal plane, while both proximal and distal muscles showed large contributions in the frontal plane. Along with the stance-limb muscles, the swing-limb muscles also exhibited considerable contribution. The gluteus medius was found to be an important individual frontal-plane control muscle; enhancing its function in pathologies could ameliorate gait by attenuating trunk sway. In addition, since gravity appreciably accelerated the trunk in the frontal plane, it may engender excessive trunk sway in pathologies.

1. Introduction

Impaired control of the trunk during walking is associated with aging and many movement disorders. This has been demonstrated as an increased motion of the trunk, for example, in the elderly (Goutier et al., 2010), in Parkinson's disease (Cole et al., 2010; Adkin et al., 2005), myotonia congenita (Horlings et al., 2009), spinocerebellar ataxia (Van de Warrenburg et al., 2005), and multiple sclerosis (Spain et al., 2012). Control of the trunk is physiologically related to control of the overall balance as follows: it reduces the acceleration of the head, which allows stabilization of the optic flow, a more effective processing of the vestibular system signals, and a consequent control of balance (e.g., Cappozzo, 1981; Mazza et al., 2008). From a mechanical point of view, the trunk is massive and its elevation is high, so it stores a large amount of gravitational potential energy. Thus, the trunk must be controlled in order to avert transformation of the potential energy to kinetic energy, that is, to avert falling. Therefore, the aforementioned impaired control of the trunk is a concern since it physiologically and mechanically increases the fall risk. More specifically, stability of the trunk during walking has been reported to be associated with fall history (Toebes et al., 2012).

Falls are the leading cause of unintentional-injury deaths (Korhonen et al., 2011). In the elderly, a fall is involved in 96% of hip fractures (Norton et al., 1997) which are a significant economical burden (Bessette et al., 2012). Falls occur most commonly while walking (e.g., Bentley and Haslam, 1998; Li et al., 2006), so investigation of fall-preventive control mechanisms of walking is of great importance.

Simulation allows quantification of the effect of muscle force on movement generation. Analyzing such contributions of individual muscles to the angular acceleration of the trunk is key to understanding how the trunk is controlled during walking. Studying the unimpaired control of the trunk reveals characteristics of good control. These characteristics can be pursued in the rehabilitation of impaired control. Chvatal and Ting (2012) have demonstrated that muscles recruited in normal walking are also recruited in atypical phases of gait, accounting for both anticipatory gait modifications

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http://dx.doi.org/10.1016/j.jbiomech.2014.04.052
0021-9290/

prior to perturbations and reactive feedback responses to perturbations. In the light of this, muscles that contribute to trunk control in normal walking may also contribute to trunk control under perturbation circumstances, attempting to prevent an impending fall.

In spite of its importance, muscle contributions to the angular acceleration of the trunk have not been addressed so far. Allen and Neptune (2012) have investigated muscle contributions to trunk energetics. Such an analysis does not, however, reveal the direction in which muscles accelerate the trunk. By studying one subject, Nott et al. (2010) have reported that all joint moments significantly contribute to the angular acceleration of the trunk. In addition to not including muscles, however, Nott et al. (2010) have modeled the pelvis and trunk as one segment, which may not accurately reflect the degrees of freedom between the pelvis and trunk. Compared to a model of separate pelvis and trunk segments, such a simplification has indeed previously been shown to have an impact on the role of the stance-limb hip moment, and on the magnitude of the joint-moment contributions to trunk energetics (Patel et al., 2007). Nevertheless, the study by Nott et al. (2010) serves as a baseline for the present study.

This study investigated individual lower-limb and trunk-muscle contributions to the angular acceleration of the trunk in normal walking of young healthy subjects, using muscle-driven simulations. A musculoskeletal model with separate pelvis and trunk segments and 92 muscle-tendon actuators was used. The objective was to identify the muscles that make the largest contributions to the angular acceleration of the trunk in the sagittal and frontal planes in different parts of the gait cycle. Since all joint moments have previously been reported to significantly contribute to the angular acceleration of the trunk (Nott et al., 2010), it was hypothesized that muscles at all joints significantly contribute to this acceleration in both the sagittal and frontal planes.

2. Methods

In this study, an induced acceleration analysis was performed so as to examine walking trials, as described in Section 2.2. To this end, previously generated simulations by John et al. (2012), which have improved on the simulations by Liu et al. (2008), were used. These simulations are described in Section 2.1.

2.1. Musculoskeletal model and simulations

Liu et al. (2008) created subject-specific simulations of walking using the OpenSim software (Delp et al., 2007). Both the software and the simulations are freely available at simtk.org. Experimental gait data included kinematics, ground reaction forces, and electromyographic (EMG) recordings (Schwartz et al., 2008). For the present analysis, the simulations of seven healthy subjects walking at their free speed were studied further. Their ages ranged from 7.0 to 18.0 years with a mean of 12.9 years. Walking speeds ranged from 1.0 to 1.3 m/s with a mean of 1.2 m/s.

The procedures for creating and testing the simulations have been described in detail by Liu et al. (2008) and John et al. (2012). Briefly, a generic musculoskeletal model, lower extremity from Delp et al. (1990) and the trunk from Anderson and Pandy (1999), with 23 degrees of freedom and 92 muscle-tendon actuators was scaled to match each subject's anthropometry. In this musculoskeletal model, a ball-socket joint models the degrees of freedom between the pelvis and trunk. Erector-spinae, external-oblique, and internal-oblique muscles cross this joint. Abbreviations for the reported muscles below are described in Table 1. The subtalar and metatarsophalangeal joints were locked at neutral anatomical angles. Dynamic inconsistency between the measured ground-reaction-force data and the model kinematics was reduced with the residual reduction algorithm by applying residual forces and torques to the pelvis of the model, and adjusting the model's mass properties and kinematics (Delp et al., 2007). Computed muscle control was used to estimate the muscle excitations that, in concert with the ground reaction forces, generated a forward dynamic simulation of each subject's gait pattern (Thelen et al., 2003; Thelen and Anderson, 2006). Computed muscle control was run with constraints on the simulated muscle activation so as to match the available EMG patterns.

The accuracy of the simulations has previously been tested in terms of lower-limb joint angles, joint moments, and muscle activations (Liu et al., 2008). Here, kinematics of the trunk and activations of the trunk muscles were tested. Experimental and simulated kinematics of the trunk for one subject are shown in Fig. 1. Among all subjects, the maximum absolute difference between the

Table 1
Descriptions of the abbreviations for the muscles reported.

Abbreviation	Description
ERCSPN	Erector spinae
EXTOBL	External oblique
INTOBL	Internal oblique
GMAX(a, im, p)	Gluteus maximus (anterior, intermedius, posterior)
GMED(a, im, p)	Gluteus medius (anterior, intermedius, posterior)
ADDL	Adductor longus
ADDM(s, m, i)	Adductor magnus (superior, medius, inferior)
SMEM	Semimembranosus
STEN	Semitendinosus
BFlh	Biceps femoris long head
RF	Rectus femoris
VASLAT	Vastus lateralis
SOL	Soleus
MEDGAS	Medial gastrocnemius
TIBANT	Tibialis anterior

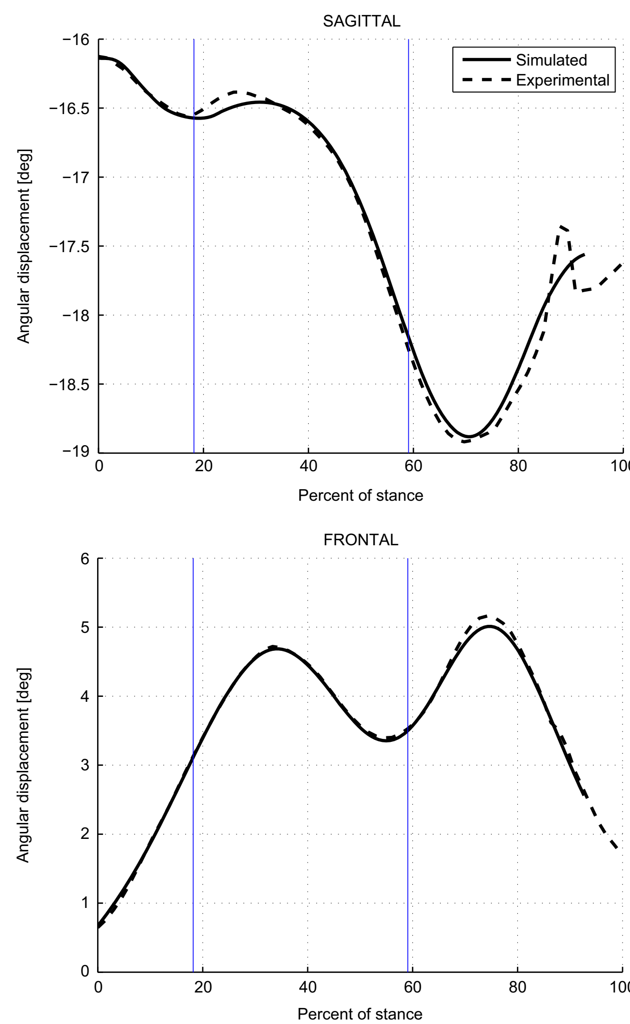

Fig. 1. Comparison of the experimental and simulated kinematics of the trunk during a right-leg stance for one subject. In the sagittal plane, positive direction corresponds to rotation where the upper trunk moves backward. In the frontal plane, positive direction corresponds to rotation where the upper trunk moves rightward. The studied phases (initial double support, and early and late single supports) are separated by the vertical solid lines.

experimental and simulated angular displacements of the trunk was smaller than 1.1 and 1.4° in the sagittal and frontal planes, respectively, showing that the simulations reproduced well the experimental kinematics of the trunk. In the simulations, internal and external obliques were active throughout the gait cycle, which is in agreement with the measurements by Waters and Morris (1972). In addition, the sum of simulated activations of back and abdominal muscles peaked at the contralateral early and late stances, as do also the EMG data of Anders et al. (2007).

2.2. Induced acceleration analysis

An induced acceleration analysis (Hamner et al., 2010; Riley and Kerrigan, 1999; Steele et al., 2013) was employed so as to compute the contributions of individual muscles to the angular acceleration of the trunk in the sagittal and frontal planes. This analysis was performed separately at each time point of the subjects' gait simulations. The analysis solves the model's equations of motion so as to calculate the angular acceleration of the trunk induced by each force (muscle force, gravity, and velocity-related forces). Contact-constraint equations enforced a rolling-without-slipping condition so as to model the foot-floor contact (Hamner et al., 2010).

Since the data did not include the entire single-support phases, assuming symmetry, the corresponding parts of previous single supports were used so as to analyze the entire gait cycle. The induced-acceleration data of each force were divided into positive and negative parts. These parts were integrated with respect to percentage of the gait phase. Such an integral is equal to the average of the induced acceleration during the gait phase. The studied gait phases included initial double support, and early and late single supports corresponding to the first and second halves of the single-support phase, i.e., the single-limb stance, respectively. Mean and standard deviations of each force's contribution (i.e., integral of the induced acceleration) across the subjects were computed for each phase. Forces that induced angular accelerations larger than 15 percent of that of the largest contributor in each phase were included for further analysis so as to only report forces that contribute more than the residuals. One of the eight subjects of the original data of Liu et al. (2008) was excluded because of significant residuals. For comparison to previous studies, the induced-acceleration analysis was also conducted so as to identify the contributions of net joint moments to the angular acceleration of the trunk.

3. Results

The induced angular acceleration curves for one subject, illustrated in Fig. 2, exemplify the joint moment and gravity contributions to the angular acceleration of the trunk. In the sagittal plane, the hip-flexion/extension and lumbar-flexion/extension moments had the largest contributions, whereas the knee-flexion/extension and ankle-plantar/dorsiflexon moments played a minor role. In the frontal plane, the majority of the joint moments made considerable contributions. For this same subject, induced angular acceleration curves for a few selected muscles are shown in Fig. 3. Like the joint moments, proximal muscles were of the greatest importance in the sagittal plane, while both proximal and distal muscles contributed to the angular acceleration of the trunk in the frontal plane.

Figs. 4 and 5 depict the group means and standard deviations of muscles' and gravity's contributions to the angular acceleration of the trunk in the sagittal and frontal planes, respectively. During the initial double support, in the sagittal plane, the back and hip-flexor muscles accelerated the trunk backward while the hamstring, abdominal, and gluteus muscles accelerated it forward. In the frontal plane, the ipsilateral and contralateral abdominal and back muscles induced the largest ipsilateral and contralateral angular acceleration, respectively. In the sagittal plane during the single-support phase, the main contributors had the same role as during the double-support phase. Contributions of individual muscles, however, varied over time. For example, contributions of the iliacus and psoas of the swing limb decreased from the initial double-support phase to the swing phase. Further, the hamstring muscle group displayed an increasing contribution from the early-swing phase to the late-swing phase. There was time evolution in the muscle contributions in the frontal plane as well. For example, the gluteus medius of the support limb exhibited a

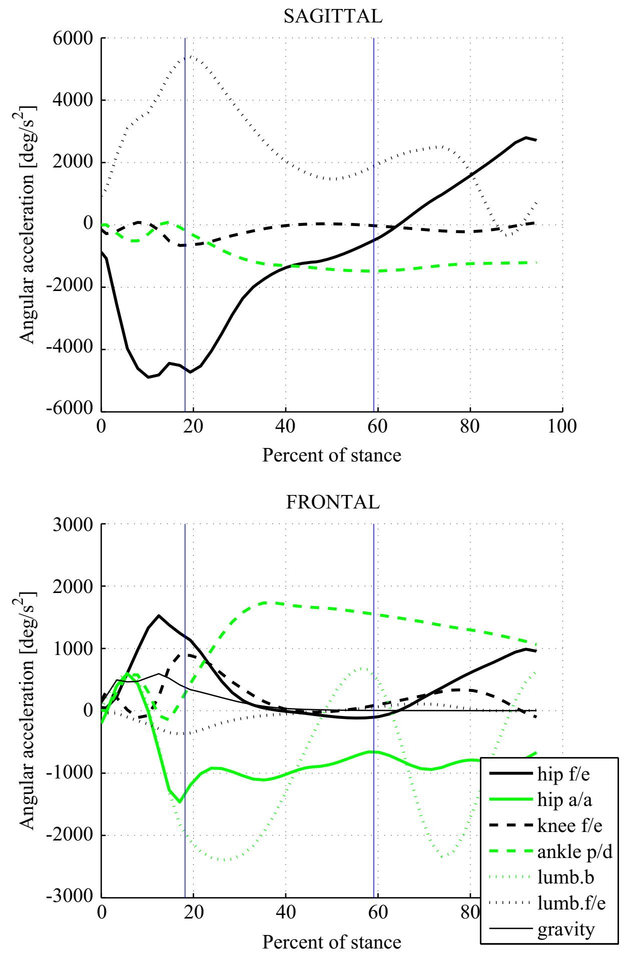

Fig. 2. Contributions of the stance-limb joint moments and gravity to the angular acceleration of the trunk during a right-leg stance for one subject. In the sagittal plane, positive direction corresponds to rotation where the upper trunk moves backward. In the frontal plane, positive direction corresponds to rotation where the upper trunk moves rightward. The studied phases (initial double support, and early and late single supports) are separated by the vertical solid lines. In the sagittal plane, the hip-flexion/extension and lumbar-flexion/extension moments made the largest contributions, whereas the knee-flexion/extension and ankle-plantar/dorsiflexon moments played a minor role. In the frontal plane, the hip-flexion/extension-, hip-abduction/adduction-, ankle-plantar/dorsiflexion-, knee-flexion/extension- and lumbar-bending moments were all important.

gradually increasing contribution from the initial double-support phase to the late single-support phase. Also, the soleus and medial gastrocnemius of the support limb had a significant contribution only during the single-support phase. Such time evolutions resulted from changes in both the muscle forces and muscles' potentials to accelerate the trunk. Gravity showed considerable contribution only in the frontal plane during the double-support phase. The high errorbars in Figs. 4 and 5 reflect high inter-subject variability.

4. Discussion

This study was conducted to identify what lower-limb, abdominal, and back muscles make the largest contributions to the angular acceleration of the trunk during walking, at different parts of the gait cycle, in the sagittal and frontal planes. Young healthy subjects walking at their free speed were studied by means of 3D muscle-driven simulations. The results showed that, in the sagittal plane, roughly, the more distal the muscles were to the trunk, the less they contributed to its angular acceleration. In contrast to this,

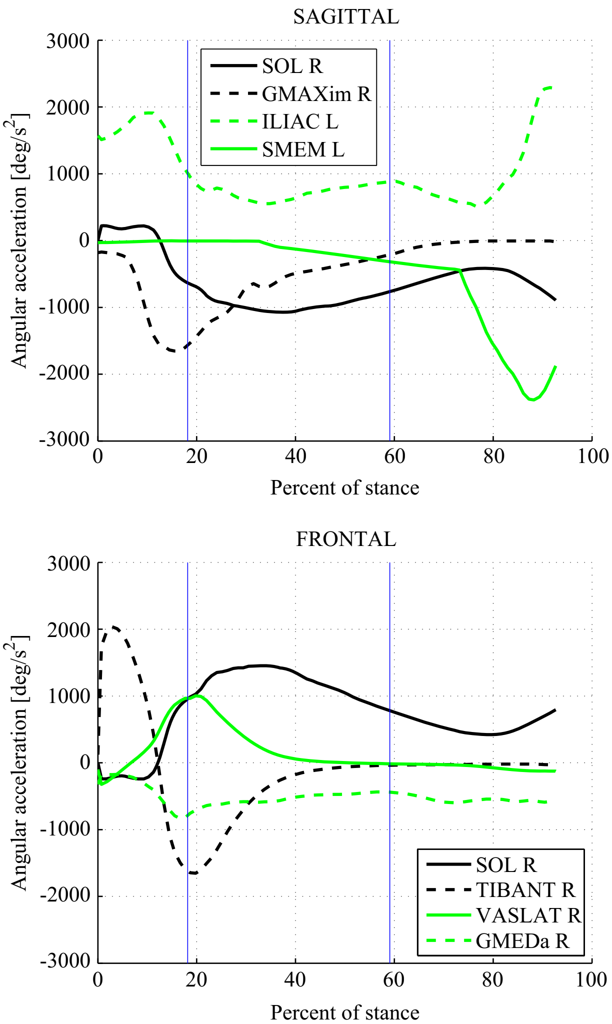

Fig. 3. An example of contributions of individual muscles to the angular acceleration of the trunk during a right-leg stance for one subject. In the sagittal plane, positive direction corresponds to rotation where the upper trunk moves backward. In the frontal plane, positive direction corresponds to rotation where the upper trunk moves rightward. The studied phases (initial double support, and early and late single supports) are separated by the vertical solid lines. In the sagittal plane, proximal muscles were of the greatest importance, while distal muscles were of great importance, too, in the frontal plane.

in the frontal plane both proximal and distal muscles exhibited large contributions. The abdominal and back muscles were of great importance throughout the gait cycle in both planes of rotation, while the lower-limb muscles displayed changes in their contributions depending on the phase of the gait cycle. In both planes, the swing-limb muscles also exhibited an appreciable contribution. There was comparatively high inter-subject variability, which made it impossible to assess the exact order of importance of the individual muscles in general. Nevertheless, a group of important muscles, such as the hip-flexors, -extensors and -abductors, abdominal, and back muscles, was identified. For comparison, muscle contributions to the angular acceleration of the trunk in the frontal plane were reminiscent of muscle contributions to the medio-lateral (John et al., 2012; Pandy et al., 2010) and vertical (Liu et al., 2008) accelerations of the center of mass. In contrast, muscle contributions to the angular acceleration of the trunk in the sagittal plane differed from muscle contributions to the fore-aft and vertical accelerations of the center of mass (Liu et al., 2008).

It has been previously reported that in the sagittal plane, all joint moments significantly contribute to the angular acceleration of the trunk (Nott et al., 2010). This result was not supported by the present findings. Such a difference between these two studies likely stemmed from the fact that, in the study by Nott et al. (2010), the model used had no degrees of freedom between the trunk and pelvis, i.e., the trunk and pelvis were a combined segment. This made the distal lower-limb segments more strongly coupled with the trunk and, consequently, led to larger contributions of the distal lower-limb joint moments to the angular acceleration of the trunk. Also, because of this simplification, the moment between the trunk and pelvis was ignored, which was here shown to be the largest contributor. In the frontal plane, the results of these two studies were similar in the sense that all the sagittal-plane lower-limb joint moments significantly contributed to the angular acceleration of the trunk. This is explicable as follows; the sagittal-plane joint moments cause reaction forces that have a moment about the trunk in the frontal plane, and these reaction forces are not dependent on the degrees of freedom of the model. Due to the difference between the models, the meaning of the trunk is not the same in these two studies, so the results cannot be compared in detail. There is currently no gold standard against which to evaluate the accuracy of the induced-acceleration-analysis predictions; however, modeling the pelvis and the trunk as separate segments is closer to the anatomical architecture of the body (Patel et al., 2007).

Along with the abdominal and back muscles, the gluteus medius was an important muscle in opposing the trunk sway in the frontal plane. Thus, enhancing the function of the gluteus medius would likely ameliorate the gait of many pathologies by attenuating trunk sway. In fact, hip abductor weakness has shown an association with the ipsilateral trunk sway in patients with cerebral palsy (Krautwurst et al., 2013), and in patients with lumbo-sacral myelomenginocele (Gutierrez et al., 2003). Soleus and medial gastrocnemius of the stance limb rotated the trunk such that the upper trunk was accelerated ipsilaterally, i.e., away from the stance limb. This should be taken into consideration when designing rehabilitation management; increased ankle-plantarflexor forces are advantageous as they help propel forward, but are disadvantageous as they destabilize the trunk, so all movement tasks should be considered as a whole. Gravity displayed considerable contribution in the frontal plane during double support. It thus may be that pathologies cannot resist such effect of gravity, which may engender the excessive trunk sway among pathologies. In other words, the greater the induced angular acceleration by gravity during double support the more difficult it becomes for muscles like the gluteus medius to slow down the motion during single support, and the more likely that the trunk sway is greater.

From a stability perspective, the contribution of the swing-limb muscles to the angular acceleration of the trunk is meaningful as it implies that the swing limb can stabilize or destabilize the trunk. For example, the swing limb can be moved quickly in response to a perturbation to stabilize the trunk, but the strong coupling between the swing limb and the trunk makes the trunk sensitive to external perturbations that are directed to the swing limb, e.g., tripping. These issues are important when considering the fall risk.

It must be noted that the trunk model used here is rather simple; the spine is modeled by a ball-and-socket joint, and only a few trunk muscles are included (i.e., erector spinae and internal and external oblique muscles). A more comprehensive spine model could have affected the results presented because of a higher number of degrees of freedom. The additional degrees of freedom would have, in turn, complicated the analysis (Christophy et al., 2012). On the other hand, several studies indicate that quite simple measures of the trunk motion can distinguish pathologies from normal individuals (e.g., Goutier et al., 2010; Spain et al., 2012). This suggests that the simple model used here is capable of capturing such differences. As for the trunk muscles, the rectus

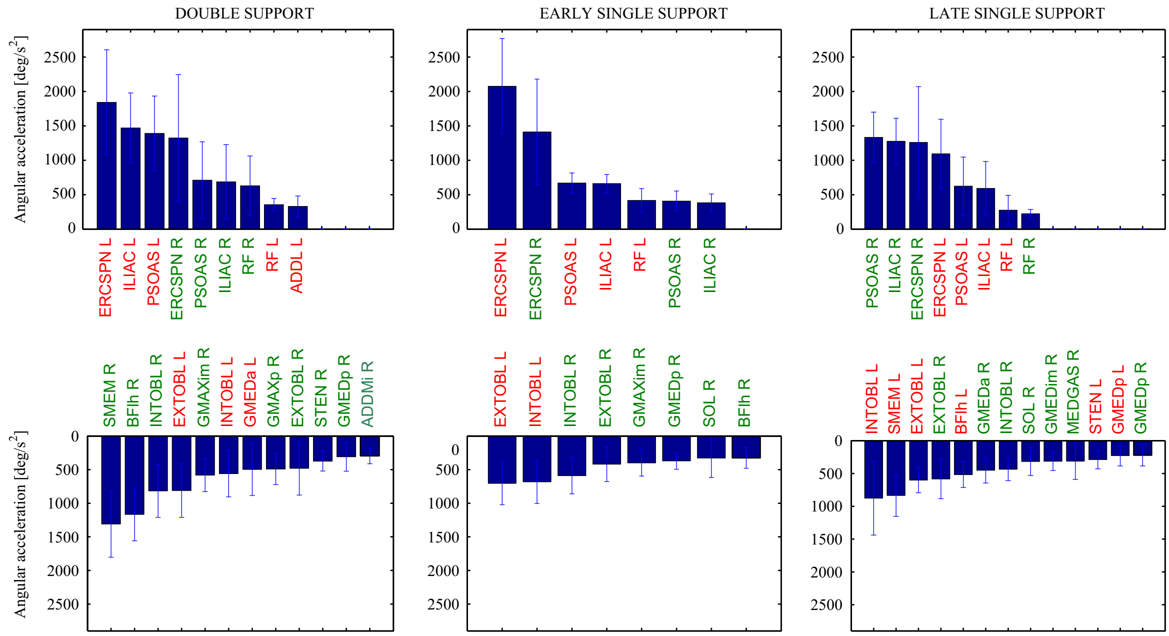

Fig. 4. Results for the sagittal plane. The group means and standard deviations of individual muscles' and gravity's contributions to the angular acceleration of the trunk in different phases of the gait cycle. The means are indicated by bars and the standard deviations by error bars. R and green color refer to the leading leg (right leg) and L and red color refer to the trailing leg (left leg). Positive direction corresponds to rotation where the upper trunk moves backward. (For interpretation of the references to color in this figure caption, the reader is referred to the web version of this article.)

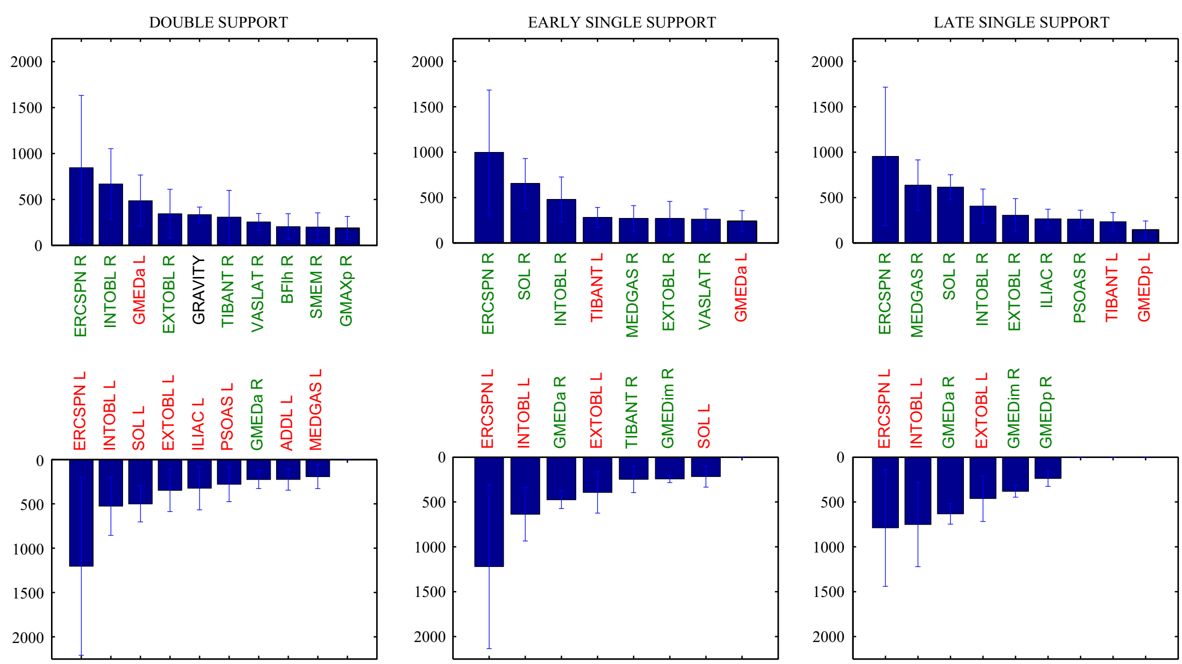

Fig. 5. Results for the frontal plane. The group means and standard deviations of individual muscles' and gravity's contributions to the angular acceleration of the trunk in different phases of the gait cycle. The means are indicated by bars and the standard deviations by error bars. R and green color refer to the leading leg (right leg) and L and red color refer to the trailing leg (left leg). Positive direction corresponds to rotation where the upper trunk moves rightward.(For interpretation of the references to color in this figure caption, the reader is referred to the web version of this article.)

abdominis, for example, would have likely made a large contribution, but it was not included in the model. Thus, the internal and external oblique muscles' contributions reported must be interpreted as the net effect of all the abdominal muscles. In other words, this study demonstrated quite an intuitive fact that the abdominal and back muscles are important, but their exact individual contributions remained unclear. Also, the paraspinal muscles have been shown to play a key role in minimizing the head motion by controlling the trunk (Prince et al., 1994). Overall, the strength of this study lies in identifying lower-limb muscle contributions in detail.

Although the musculoskeletal model used does not include the arms, the residual forces and moments applied to the model are intended, in part, to represent the effects of the arms. Arm motion certainly contributes to the angular acceleration of the trunk. Nevertheless, contribution of the arms can be considered to be smaller than that of the muscles that were reported here, since only muscles that contributed more than the residuals were reported. In general, as far as arms swing symmetrically, their angular momenta cancel each other in the sagittal and frontal planes and, consequently, they do not affect the trunk motion. In the transverse plane, there is no such cancelation, but the transverse plane was not analyzed here. Also, the subtalar joint was locked, which might have an impact on the results. This impact is expected to be small since Pandy et al. (2010) and John et al. (2012) have reported similar findings although in the former study the subtalar joint was not locked while in the latter one it was.

The results should also be interpreted in the light of the fact that, in the induced acceleration analysis, the estimates of muscle contributions depend on simulated muscle forces. A direct validation of the simulated muscle forces is not possible as the actual muscle forces of a human subject cannot usually be measured. This poses a challenge for creating completely subject-specific simulations. Alteration of the muscle forces does not, however, affect the direction of the angular acceleration induced by each muscle, but it can alter the relative contributions of the muscles.

The ages of the subjects studied ranged from 7 to 18 years. Nevertheless, the subjects can be considered to be representative of adults as walking kinematics and joint moments become adult-like by age 7 (Sutherland, 1997). Moreover, kinematics, joint moments, and muscle activations of the simulations agree with measurements on adults (Liu et al., 2008).

In conclusion, in contrast to the hypothesis, muscles at all joints significantly contributed to the angular acceleration of the trunk only in the frontal plane in normal walking of young healthy subjects: the proximal muscles were more important than the distal ones in the sagittal plane, while a combination of proximal and distal muscles was recruited for the frontal-plane control.

Conflict of interest statement

None of the authors had financial or personal conflict of interest with regard to this study.

Acknowledgments

The authors thank a former member of the Neuromuscular Biomechanics Lab, at Stanford University, Chand John, for sharing his simulations.

References

Adkin, A.L., Bloem, B.R., Allum, J.H., 2005. Trunk sway measurements during stance and gait tasks in Parkinson's disease. Gait Posture 22, 240–249.
Allen, J.L., Neptune, R.R., 2012. Three-dimensional modular control of human walking. J. Biomech. 45, 2157–2163.
Anders, C., Wagner, H., Puta, C., Grassme, R., Petrovitch, A., Scholle, H.C., 2007. Trunk muscle activation patterns during walking at different speeds. J. Electromyogr. Kinesiol. 17, 245–252.
Anderson, F.C., Pandy, M.G., 1999. A dynamic optimization solution for vertical jumping in three dimensions. Comput. Methods Biomech. Biomed. Eng. 2, 201–231.
Bentley, T.A., Haslam, R.A., 1998. Slip, trip, and fall accidents occurring during the delivery of mail. Ergonomics 41, 1859–1872.
Bessette, L., Jean, S., Lapointe-Garant, M.P., Belzile, E.L., Davison, K.S., Ste-Marie, L.G., Brown, J.P., 2012. Direct medical costs attributable to peripheral fractures in Canadian post-menopausal women. Osteoporos Int. 23, 1757–1768.
Cappozzo, A., 1981. Analysis of the linear displacement of the head and trunk during walking at different speeds. J. Biomech. 14, 411–425.
Christophy, M., Faruk Senan, N.A., Lotz, J.C., O'Reilly, O.M., 2012. A musculoskeletal model for the lumbar spine. Biomech. Model. Mechanobiol. 11, 19–34.
Chvatal, S.A., Ting, L.H., 2012. Voluntary and reactive recruitment of locomotor muscle synergies during perturbed walking. J. Neurosci. 32, 12237–12250.
Cole, M.H., Silburn, P.A., Wood, J.M., Worringham, C.J., Kerr, G.K., 2010. Falls in Parkinson's disease: kinematic evidence for impaired head and trunk control. Mov. Disord. 24, 2369–2378.
Delp, S.L., Loan, J.P., Hoy, M.G., Zajac, F.E., Topp, E.L., Rosen, J.M., 1990. An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. IEEE Trans. Biomed. Eng. 37, 757–767.
Delp, S.L., Anderson, F.C., Arnold, A.S., Loan, P., Habib, A., John, C.T., Guendelman, E., Thelen, D.G., 2007. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans. Biomed. Eng. 54, 1940–1950.
Goutier, K.M., Jansen, S.L., Horlings, C.G., Kung, U.M., Allen, J.H., 2010. The influence of walking speed and gender on trunk sway for the healthy young and older adult. Age Ageing 39, 647–650.
Gutierrez, E.M., Bartonek, A., Haglund-Akerlind, Y., Saraste, H., 2003. Characteristic gait dynamics in persons with lumbosacral myelomenginocele. Gait Posture 18, 170–177.
Hamner, S.R., Seth, A., Delp, S.L., 2010. Muscle contributions to propulsion and support during running. J. Biomech. 43, 2709–2716.
Horlings, C.G., Drost, G., Bloem, B.R., Trip, J., Pieterse, A.J., van Engelen, B.G., Allum, J.H., 2009. Trunk sway analysis to quantify the warm-up phenomenon in myotonia congenita patients. J. Neurol. Neurosurg. Psychiatr. 80, 207–212.
John, C.T., Schwartz, M.H., Delp, S.L., 2012. Contributions of muscles to mediolateral ground reaction force over a range of walking speeds. J. Biomech. 45, 2438–2443.
Korhonen, N., Niemi, S., Parkkari, J., Palvanen, M., Kannus, P., 2011. Unintentional injury deaths among adult Finss in 1971–2008. Injury 42, 885–888.
Krautwurst, B.K., Wolf, S.I., Heitzmann, D.W., Gantz, S., Braatz, F., Dreher, T., 2013. The influence of hip abductor weakness on frontal plane motion of the trunk and pelvis in patients with cerebral palsy. Res. Dev. Disabil. 34, 1198–1203.
Li, W., Keegan, T.H., Stemfeld, B., Sidney, S., Quesenberry Jr., C.P., Kelsey, J.L., 2006. Outdoor falls among middle-aged and older adults: a neglected public health problem. Am. J. Publ. Health 96, 1192–2000.
Liu, M.Q., Anderson, F.C., Schwartz, M.H., Delp, S.L., 2008. Muscle contributions to support and progression over a range of walking speeds. J. Biomech. 41, 3243–3252.
Mazza, C., Iosa, M., Pecoraro, F., Cappozzo, A., 2008. Control of the upper body accelerations in young and elderly women during level walking. J. NeuroEng. Rehabil., 5.
Norton, R., Campbell, A.J., Lee Joe, T., Robinson, E., Butler, M., 1997. Circumstances of falls resulting in hip fractures among older people. J. Am. Geriatr. Soc. 45, 1108–1112.
Nott, C.R., Zajac, F.E., Neptune, R.R., Kautz, S.A., 2010. All joint moments significantly contribute to trunk angular acceleration. J. Biomech. 43, 2648–2652.
Pandy, M.G., Lin, Y.C., Kim, H.J., 2010. Muscle coordination of mediolateral balance in normal walking. J. Biomech. 43, 2055–2064.
Patel, M., Talay, M., Ounpuu, S., 2007. The impact of adding trunk motion to the interpretation of the role of joint moments during normal walking. J. Biomech. 40, 3563–3569.
Prince, F., Winter, D.A., Stergio, P., Walt, S.E., 1994. Anticipatory control of upper body balance during human locomotion. Gait Posture 2, 19–25.
Riley, P.O., Kerrigan, D.C., 1999. Kinetics of stiff-legged gait: induced acceleration analysis. IEEE Trans. Rehabil. Eng. 7, 420–426.
Schwartz, M.H., Rozumalski, A., Trost, J.P., 2008. The effect of walking speed on the gait of typically developing children. J. Biomech. 41, 1639–1650.
Spain, R.I., St George, R.J., Salarian, A., Mancini, M., Wagner, J.M., Horak, F.B., Bourdette, D., 2012. Body-worn motion sensors detect balance and gait defects in people with multiple sclerosis who have normal walking speed. Gait Posture 35, 573–578.
Steele, K.M., Seth, A., Hicks, J.L., Schwartz, M.S., Delp, S.L., 2013. Muscle contributions to vertical and fore-aft accelerations are altered in subjects with crouch gait. Gait Posture 38, 86–91.
Sutherland, D., 1997. The development of mature gait. Gait Posture 6, 163–170.
Thelen, D.G., Anderson, F.C., Delp, S.L., 2003. Generating dynamic simulations of movement using computed muscle control. J. Biomech. 36, 321–328.
Thelen, D.G., Anderson, F.C., 2006. Using computed muscle control to generate forward dynamic simulation of human walking from experimental data. J. Biomech. 39, 1107–1115.
Toebes, M.J., Hoozemans, M.J., Furrer, R., Dekker, J., van Dieen, J.H., 2012. Local dynamic stability and variability of gait are associated with fall history in elderly subjects. Gait Posture 36, 527–531.
Van de Warrenburg, B.P., Bakker, M., Kremer, B.P., Bloem, B.T., Allum, J.H., 2005. Trunk sway in patients with spinocerebellar ataxia. Mov. Disord. 20, 1006–1013.
Waters, R.L., Morris, J.M., 1972. Electrical activity of muscles of the trunk during walking. J. Anat. 111, 191–199.