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Force Adaptation in Human Walking with Symmetrically Applied Downward Forces on the Pelvis

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Abstract— **The application of external constraints and/or applied forces during movement can lead to reactive as well as adaptive changes in human motion. Previous researches in the literature have usually focused on adaptation in human kinematics when external forces were applied using exoskeletons during the swing phase of gait. This work aims to study adaptation in human walking when externally applied forces are present on the pelvis both during the swing and stance phases of the gait. A novel tethered pelvic assist device (TPAD) was used to passively apply downward forces on the human pelvis while walking. During the experiment, healthy subjects walked on a treadmill at a constant speed while their kinematics and foot pressure data were recorded. Data analysis revealed that the healthy subjects exhibited both reactive as well as adaptive changes in their gait parameters. The immediate response of the subjects was to increase their hip flexion to clear their foot off the ground as they were unable to lift their pelvis to their usual height during normal motion. Seven out of eight subjects in the study resisted the downward forces to move their pelvis up. Eventually, they reached a level of downward force that they could sustain over the training. This adaptation to the downward force was reflected in the heel peak pressure values during the cycles of the gait. On removing the tethers, aftereffects in heel peak pressure during the gait cycles were observed.**

I. Introduction

The human nervous system is capable of modifying motor commands in response to alteration in movement conditions [1]-[10]. During walking, such external alterations generate errors in movement kinematics. Humans minimize these errors by recalibrating motor commands [11]. The changes in response to the situation can be either adaptive or merely immediate [3], [6]. Humans walking on a split belt treadmill modify the stance timing immediately in response to the setting of the belts at different speeds. This lack of adaptation in stance time differs from the adaptive response demonstrated in changes of step length, even after the two belts are once again returned to the same speed [6].

Most studies on adaptation of human gait involve the use of robot exoskeletons for generation of external force fields to perturb human walking during the swing phase of the gait. In [3], [4], the Lokomat was used to provide velocity-dependent resistance against hip and knee movements. An assist-as-needed paradigm was implemented at the hip and the knee using ALEX in response to errors in the position of the ankle with respect to its target template [8], [9]. In [12], [13], the gait rehabilitation robot LOPES was used to apply forces at the ankle based on a spring-damper model. Similarly, cables were used to provide assistance/resistance to the leg during swing [10]. The perturbations with these devices were applied during the swing phase of gait. However, the human nervous system is flexible to accommodate changes in both the swing and stance phases of walking. It is important to point out that reported studies on the effect of externally applied forces during walking have been limited to those during swing.

Pelvic motion plays an important role in gait as it assists forward propulsion of the body by transferring forces from the lower extremity to the trunk [14]. It is also crucial in assisting swing initiation and in modulating the vertical displacement of the body's center of mass. This helps in reducing energy consumption during walking [15]. Therefore, strategies for adaptation in force at the pelvis have the potential to influence both the stance and swing phases of walking.

Tethered Pelvic Assist Device (TPAD) was designed to assist in understanding issues of adaptation of forces applied during a gait cycle. TPAD is a passive system that consists solely of springs and cables. The springs within TPAD have the capability of storing mechanical energy. TPAD provides pelvic support in the form of applied forces in any direction, including the direction of gravity. Additionally, its tethers can also be configured to apply asymmetric forces on the pelvis. Because of its passive nature, TPAD is safe and cost effective when compared to active devices such as the PAM [16] and KineAssist [17]. A computational method for designing a passive pelvis device was reported in [18]. An initial feasibility study using the TPAD with healthy subjects was reported in [19].

The objective of this work is to study the adaptation in human walking in response to externally applied forces on the pelvis during the swing and stance phases. We hypothesize that downward forces on the pelvis will constrain the pelvic range of motion and will shorten the distance between the pelvis and the ground. In order to walk in a more erect posture, subjects will need to work against the applied downward forces. The elastic nature of the tether will allow them to do so until they reach a balance between the extent of the erect posture and downward forces that they can sustain during walking. As a result, training with these forces for 15 minutes will induce changes in their gait kinetics showing aftereffects.

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II. Method

A. Experimental Setup

In order to apply external forces during the swing and stance phases of walking, the Tethered Pelvic Assist Device (TPAD) was used [19]. As shown in Fig. 1, one end of each tether was attached to a hip brace while the other to an inertially fixed frame around the subject. Each elastic tether consists of a spring connected in series with a cable. During experiments, subjects folded their arms and rested these on their chest to avoid potential interference with the elastic tethers. In the current study, the human pelvis was symmetrically loaded downwards as shown in Fig. 1. Four tethers were directed downwards towards the floor while two were connected to the top of the frame. The purpose of tethers connected to the top frame was to ensure that the subjects felt balanced while walking. In the experiment, a spring of 4.04 N/mm stiffness was used for each tether. An initial tension of 90-100 N was set in each tether while the subject stood still and straight on the treadmill. These values were chosen so that subjects could safely balance and walk on the treadmill at 2.5 mph for an extended period of time.

B. Experimental Protocol

Eight healthy male subjects, age range 24-31 years (mean age: 27 yrs and SD: 2.33 yr), mean weight 76.12 kg (SD: 12.38 kg) participated in the experiment. The study was approved by the University of Delaware Internal Review Board and all subjects provided written consent. The study involved the following training and evaluation sessions.

*Session 1 (**Baseline**):* The subject walked on the treadmill at a constant speed of 2.5 mph for two minutes. Data collected during this session was treated as reference and referred to as BL in this paper.

*Session 2 (**Training**):* The tethers were attached to the hip brace and a constant tension value was adjusted in each tether. The subject then walked for fifteen minutes with the tethers attached at the same treadmill speed. During this session, data was recorded five times for one minute duration, i.e., at start, 3^{rd}, 6^{th}, 9^{th} and 12^{th} min. These data collection instances were referred to as ET (early training), T2, T3, T4 and LT (late training).

*Session 3 (**Post training**):* The tethers were removed, and immediately, the subject was asked to walk for another nine minutes at the same treadmill speed. Data was recorded at start, 3^{rd} and 6^{th} min and these instances were referred to as EPT (early post training), PT2 and LPT (late post training).

C. Data Processing

Data was collected from the motion capture system and the tension sensor system at 100 Hz. However, the micro-controller for foot pressure recorded data at 35 Hz. Gait events, such as toe-off and heel strike, were determined based on toe and heel markers' position with respect to sacrum marker as illustrated in [22]. Further, as the timing information was synchronized by an external signal among three data collection stations, the heel strike events were used to segment the pressure and the tension data into gait cycles.

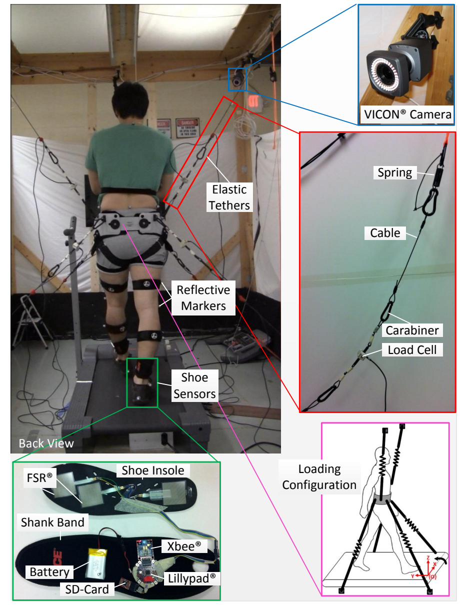


Fig. 1. Experimental setup of TPAD in the loading configuration - Motion capture system (from VICON®) was used to track the reflective markers. Load cells (MLP-200 from Transducer Techniques®) were used to measure the amount of tension in each tether. Shoe insoles with pressure pads (FSR®) and shank band with micro-controller (Arduino® Lillypad) were used to record the foot pressure.

Voltage readings of the pressure pads were converted to pressure units using a non-linear calibration graph provided by the manufacture.

In the experiment, the constraints were applied at the pelvis. Therefore, pelvic range of motion as well as hip, knee and ankle sagittal plane angles were compared between sessions. The tension data from each cable was used to calculate the resultant force and moment at the origin of sacrum marker on the human body. It was observed that one subject did not resist the forces and hence was excluded from the group analysis. Adaptation to the vertical force on pelvis during training directly influenced the foot pressure. Peak pressure values were compared between sessions to estimate the adaptation and aftereffects of training with force. The Kolmogorov-Smirnov test with Lilliefors correction was performed across the sessions to check the normality assumption of the data. Repeated measure ANOVA was performed to determine the statistical significance (defined as $p < 0.05$). Tukey's post-hoc honestly significant difference test was performed when a statistical significance was identified. Further, values plotted in the following section are means $\pm$ standard errors.

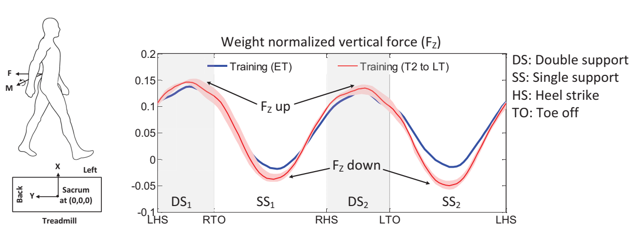
III. Results

A. External forces

The resultant force and moment due to combined effect of tethers, resolved on the sacrum marker of the human body, is shown in Fig. 2(a). These force and moment components adapt over the 15 $mins$ of training. The vertical force component (F_Z) was analyzed to understand subjects' adaptation to the applied constraints. Figure 2(b) shows the variation of vertical force for a representative subject during training. Subjects experienced negative force value (F_Z down) only during single support phases of walking, marked as SS_1 and SS_2. During the double support phases, DS_1 and DS_2, the nature of vertical force was positive (F_Z up). An increase in the F_Z down value can be seen as the training session progresses indicating that subjects were resisting the downward force. The F_Z up value remained consistent throughout the training. This increase in F_Z down value with training was shown by seven subjects. Figure 3(a) presents the group results for F_Z down and up values. Clearly, subjects adapted to the downward vertical value. There was a significant increase in F_Z down value from ET to LT and from ET to T2 ($p < 0.05$). Further, no significance was observed for F_Z up values.

B. Pelvic motion

As reported in earlier study with TPAD [19], attachment of elastic tethers reduced pelvic range of motion. In the current study, both pelvic translational and rotational motion range were constrained. The vertical pelvic motion during a gait cycle directly affects the magnitude of F_Z force. To study the pelvic vertical motion, pelvic center was calculated as the centroid of the triangle formed by three pelvis markers (right and left ASI and sacrum). Pelvic highest and lowest positions during different sessions with respect to the static trial value have been plotted in Fig. 3(b) for the group. Significant decrease in pelvic highest position was observed from BL to ET (or LT) and from EPT to LT ($p < 0.05$), but not within training or between baseline and post training. This means that the lower pelvic position during the single support phase of walking was only an immediate response to applied forces and it went back to the normal once the constraints were removed. Pelvic lowest position, in the double support phase, did not change considerably during training as seen in Fig. 3(b).

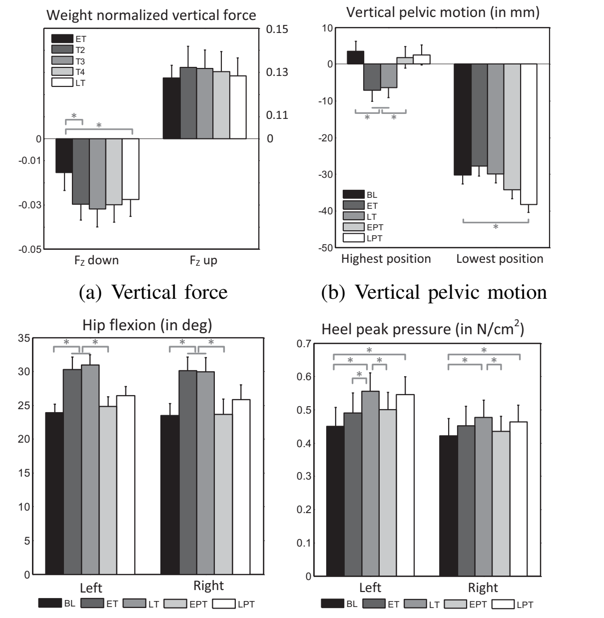


(a) (b)

Fig. 2. (a) Resultant force and moment vector, (b) Weight normalized vertical component of the resultant force for a representative subject. Increase in F_Z down value can be observed as training progresses.

C. Joint angles

From the bar graph in Fig. 3(c), it can be seen that the left as well as right hip flexion values increased from BL to ET (or LT) and from EPT to LT ($p < 0.05$), but not within training or between baseline and post training. This is similar to the trend observed for pelvic highest position and therefore the increase in hip flexion was also a reactive change.



(a) Vertical force (b) Vertical pelvic motion

(c) Hip flexion (d) Heel peak pressure

Fig. 3. Bar graphs for the group (excluding subject 7). F_Z down value adapts during training while the variation in pelvic highest position was a reactive response to the applied constraints. A reactive response was also noticed in hip flexion value during training. Heel peak pressure increased during training and showed aftereffects. '*' denotes significant difference ($p < 0.05$).

D. Foot pressure

An increase in heel pressure was observed during training as well as during post training. As seen in Fig. 3(d), the peak pressure value increased significantly from BL to LT ($p < 0.05$) for both legs. An increase in pressure value was expected during training because of the induced net external downward force (F_Z down) on the pelvis. The peak pressure value also increased from ET to LT, though significantly only for the left heel, implying adaptation in response to adaptation in F_Z down value. During the post training session, peak pressure values remained higher as compared to baseline, see Fig. 3(d). The heel peak pressure value for both legs was significantly higher during LPT compared to BL ($p < 0.05$).

IV. Discussion

The novelty of the current work is the application of external forces during both the swing and stance phases of walking. Previous work examining adaptation to external forces during walking [2], [3], [5], [7], [8], [9], [10] have only employed external forces during the swing phase of walking. Results obtained from the experiment show adaptive as well as reactive changes in the gait parameters.

In the results, we found that the amount of downward force (F_Z down) on the pelvis increased during the early part of training (ET) but stabilized after some time. At the start, subjects were unaware of the nature of applied forces. As the training session progressed, they applied resistance against the pulling forces of the elastic tethers. Inherent passivity of the setup allowed them to reach a configuration which they could sustain for the rest of the training session. In literature [3], [6], such adaptive human behaviors in response to the applied constraints are associated with the recalibration of motor commands, which cause aftereffects when the constraints are removed. In current study, aftereffects were observed in heel pressure values. Further, resisting the force was not the only strategy that subjects utilized. One subject (subject 7) did not resist the forces and showed no adaptation during walking with the applied forces. This subject was not included in the current analysis.

Results also showed that the presence of external constraints on the pelvis reduced its range of motion. Particularly, subjects were unable to lift their pelvis to the normal height during the single support phases. This led them to increase their hip flexion to clear their foot off the ground. Such immediate responses correspond to the feedback control strategies adopted by humans in response to applied perturbations, as explained in [3].

The elastic nature of TPAD ensures that the external forces are present both during the swing and stance phases of the gait cycle. One very important advantage of using such an arrangement is that the downward forces can be applied without modifying the subjects' inertia properties. In previous studies with added weight [20], [21], subjects' inertia was also modified.

V. Conclusion

This paper studied adaptation in human walking when downward forces were applied to the pelvis both during the swing and stance phases of the gait. Results showed that healthy subjects were unable to lift their pelvis to their normal height in the presence of external constraints. Therefore, in order to clear their foot off the ground, they increased their hip flexion. The group adopted a common strategy to adapt to the externally applied downward forces on the pelvis that resulted in greater foot pressure during post training. In conclusion, symmetrically applied downward forces on the pelvis led to adaptation in gait kinetics, specially, higher pressure at the heel.

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References

[1] D. S. Reisman, H. J. Block and A. J. Bastian, "Interlimb Coordination During Locomotion: What Can be Adapted and Stored", *J Neurophysiol*, vol. 94, pp. 2403–2415, 2005.

[2] J. L. Emken and D. J. Reinkensmeyer, "Robot-enhanced motor learning: accelerating internal model formation during locomotion by transient dynamic amplification", *IEEE Trans Neural Syst Rehabil Eng*, vol. 13, pp. 33–39, 2005.

[3] T. Lam, M. Anderschitz and V. Dietz, "Contribution of Feedback and Feedforward Strategies to Locomotor Adaptations", *J Neurophysiol*, vol. 95, pp. 766–773, 2006.

[4] A. Houldin, K. Luttin and T. Lam, "Locomotor adaptations and after effects to resistance during walking in individuals with spinal cord injury", *J Neurophysiol* vol. 106, pp. 247-258, 2011.

[5] J. W. Noble and S. D. Prentice, "Adaptation to unilateral change in lower limb mechanical properties during human walking", *Exp Brain Res*, vol. 169, pp. 482–495, 2006.

[6] D. S. Reisman, R. Wityk, K. Silver and A. J. Bastian, "Locomotor adaptation on a split-belt treadmill can improve walking symmetry post-stroke", *Brain*, vol. 130, pp. 1861–1872, 2007.

[7] A. Blanchette and L. J. Bouyer, "Timing-specific transfer of adapted muscle activity after walking in an elastic force field", *J Neurophysiol*, vol. 102, pp. 568–577, 2009.

[8] S. K. Banala, S. H. Kim, S. K. Agrawal and J. P. Scholz, "Robot Assisted Gait Training With Active Leg Exoskeleton (ALEX)", *IEEE Trans Neural Syst Rehab Eng*, vol. 17, No. 1, 2009.

[9] S. H. Kim, S. K. Banala, E. A. Brackbill, S. K. Agrawal, V. Krishnamoorthy and J. P. Scholz, "Robot-assisted modifications of gait in healthy individuals", *Exp Brain Res*, vol. 202, pp. 809–824, 2010.

[10] S. C. Yen, B. D. Schmit, J. M. Landry, H. Roth and M. Wu, "Locomotor adaptation to resistance during treadmill training transfers to overground walking in human SCI", *Exp Brain Res*, Nov, 2011.

[11] A. J. Bastian, "Understanding sensorimotor adaptation and learning for rehabilitation", *Current opinion in neurology*, vol. 21, 2008.

[12] E. H. F. van Asseldonk, B. Koopman, J. H. Buurke, C. D. Simons and H. van der Kooij, "Selective control of a subtask of walking in a robotic gait trainer(LOPES)", *10^{th} International Conference on Rehabilitation Robotics,*, Noordwijk, The Netherlands, June 12-15, 2007.

[13] E. H. F. van Asseldonk, B. Koopman, J. H. Buurke, C. D. Simons and H. van der Kooij, "Selective and adaptive robotic support of foot clearance for training stroke survivors with stiff knee gait", *11^{th} International Conference on Rehabilitation Robotics*, Japan, June 23-26, 2009.

[14] J. Perry, *Gait analysis: Normal and pathological function*, SLACK Inc, Thorofare, 1992.

[15] R. R. Neptune, S. A. Kautz, and F. E. Zajac, "Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking", *J Biomechanic*, vol. 34(11), 2001.

[16] D. Aoyagi et. al, "A Robot and Control Algorithm That Can Synchronously Assist in Naturalistic Motion During Body-Weight-Supported Gait Training Following Neurologic Injury", *IEEE Transactions on Neural Systems and Rehab. Engg.*, vol. 15, no. 3, September 2007.

[17] M. Oeshkin et. al, "KineAssist: A robotic over- ground gait and balance training device", *in proc. IEEE Conf. Rehab. Robot.*, pp. 241–246, 2005.

[18] A. Mokhtarian, A. Fattah and S. K. Agrawal, "A novel passive pelvic device for assistance during locomotion", *in proc. IEEE Int. Conf. Robot. Auto.*, pp. 2241–2246, 2010.

[19] V. Vashista, S. K. Mustafa and S. K. Agrawal, "Experimental Studies On The Human Gait Using A Tethered Pelvic Assist Device (T-PAD)", *IEEE ICORR*, ETH Zurich Science City, Switzerland, June 29 - July 1, 2011.

[20] C. P. McGowan, R. R. Neptune and R. Kram, "Independent effects of weight and mass on plantar flexor activity during walking: implications for their contributions to body support and forward propulsion", *Journal of applied physiology*, vol. 105, pp. 486–494, 2008.

[21] C. P. McGowan, R. Kram and R. R. Neptune, "Modulation of leg muscle function in response to altered demand for body support and forward propulsion during walking", *Journal of biomechanics*, vol. 42, pp. 850–856, 2009.

[22] J. A. Zeni Jr., J. G. Richards and J. S. Higginson, "Two simple methods for determining gait events during treadmill and overground walking using kinematics data", *Gait & Posture*, vol. 27, pp. 710–714, 2008.

[23] K. K. Patterson, W. H. Gage, D. Brooks, S. E. Black, and W. E. McIlroy, "Evaluation of gait symmetry after stroke: a comparison of current methods and recommendations for standardization", *Gait & posture*, vol. 31, no. 2, pp. 2416, Feb. 2010.