Dynamic Balancing Responses in Unilateral Transtibial Amputees Following Outward-Directed Perturbations During Slow Treadmill Walking Differ Considerably for Amputated and Non-Amputated Side

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Research

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Abstract

Background

Due to disrupted motor and proprioceptive function lower limb amputation imposes considerable challenges associated with balance and greatly increases risk of falling in case of perturbations during walking. The aim of this study was to investigate dynamic balancing responses in unilateral transtibial amputees when they were subjected to perturbing pushes to the pelvis in outward direction at the time of foot strike on non-amputated and amputated side during slow walking.

Methods

Fourteen subjects with unilateral transtibial amputation and nine control subjects participated in the study. They were subjected to perturbations that were delivered to the pelvis at the time of foot strike of either the left or right leg. We recorded trajectories of center of pressure and center of mass, durations of in-stance and stepping periods as well as ground reaction forces. Statistical analysis was performed to determine significant differences in dynamic balancing responses between control subjects and subjects with amputation when subjected to outward-directed perturbation upon entering stance phases with non-amputated or amputated side.

Results

When outward-directed perturbations were delivered at the time of foot strike of the non-amputated leg, subjects with amputation were able to modulate center of pressure and ground reaction force similarly as control subjects which indicates application of in-stance balancing strategies. On the other hand, there was a complete lack of in-stance response when perturbations were delivered when the amputated leg entered the stance phase. Subjects with amputations instead used the stepping strategy and adjusted placement of the non-amputated leg in the ensuing stance phase to make a cross-step. Such response resulted in significantly higher displacement of center of mass.

Conclusions

Results of this study suggest that due to the absence of the COP modulation mechanism, which is normally supplied by ankle motor function, people with unilateral transtibial amputation are compelled to choose the stepping strategy over in-stance strategy when they are subjected to outward-directed perturbation on the amputated side. However, the stepping response is less efficient than in-stance response. To improve their balancing responses to unexpected balance perturbation people fitted with passive transtibial prostheses should undergo perturbation-based balance training during clinical rehabilitation.

Background

Approximately half of community-living persons with lower limb amputation fall each year (Miller et al., 2001). Proper regulation of center-of-mass (COM) and whole-body angular momentum during walking is necessary to prevent falls, particularly in the frontal plane where individuals are most unstable (Miller et al., 2018). Thus, understanding the biomechanical aspects of balancing mechanisms used by people with lower-limb amputation fitted with passive prosthesis is crucial for deriving appropriate training programs that aim to improve their mobility.

Several studies have investigated balancing responses in subjects with Unilateral Transtibial Amputation (UTA) following unexpected loss of balance during standing (Bolger et al., 2014; Curtze et al., 2010). Findings of these studies suggest that UTA subjects were equally efficient in recovering from perturbation as healthy controls, as shown in the same efficiency to control COM although they have exhibited different excursions of center-of-pressure (COP) and stepping responses on the intact and the prosthetic leg. While standing enables a multitude of balancing strategies that can be shared between both limbs (Hof, 2007) this is not very likely during walking where most of the time either only the non-amputated or amputated leg is in contact with the ground. Few studies investigating balance during walking in UTA subjects used a multi-sine function to induce continuous pseudo-random movement of a walking surface alone (Hak et al., 2013; Beurskens et al. 2014) and combined with manipulation of visual field (Beurskens et al. 2014). Results of these studies have shown proactive changes in gait parameters that manifested in decreased step lengths, increased cadence and wider steps which enabled UTA subjects to handle pseudo-random perturbations in a similarly efficient fashion as control subjects (Beurskens et al. 2014, Hak et al., 2013). Proactive adaptations of gait parameters were also investigated in experiments where perturbing pushes were delivered to the pelvis in the frontal plane to mimic situations when people collide with other persons or objects in the environment. It was found that when the perturbation timing was unknown a priori no adaptations took place; thus, unexpected perturbation had to be negotiated purely through feedback mechanisms (Major et al., 2018).

Our knowledge of the dynamic balancing responses used by UTA subjects following unexpected loss of balance during walking is scarce due to the lack of perturbation studies. Additionally, previous research mainly focused on the stepping aspects of balance preservation where UTA subjects walked at treadmill speeds of 0.8 m/s (Major et al., 2018) and 1.2 m/s (Hak et al. 2013; Beurskens et al. 2014; Miller et al. 2018). However, the majority of UTA subjects at discharge typically walk at approximately 0.5 m/s (Batten et al. 2019). Several studies have applied outward- and inward-directed pushes to the pelvis of walking subjects to investigate the repertoire of dynamic balancing responses in healthy subjects (Hof et al., 2010; Matjačić et al., 2017; Vlutters et al., 2018a). While responses to inward perturbations, directed medially with respect to the leg instance, are fairly uniform across walking speeds and perturbation parameters, the responses to outward perturbations, directed laterally with respect to the leg instance, are more complex (Matjačić et al., 2019). When the walking speed is relatively high (above 0.8 m/s) responses to unexpected perturbation consist primarily of adequate placement of ensuing step(s), which has been termed as the “stepping” strategy (Bruijn and Dieen, 2018; Matjačić et al., 2019). However, when counteracting outward perturbations at lower walking speeds “in-stance” strategy plays the dominant role (Matjačić et al., 2019). “In-stance” strategy is composed of: “medio-lateral ankle strategy”, characterized

with lateral displacement of COP in the direction of perturbation; “braking strategy”, characterized with anterior displacement of COP resulting in braking of COM in the plane of progression; and “inertial strategy”, characterized with impulse-like increase of lateral component of ground-reaction-force (GRF) that opposes the action of perturbation (Matjačić et al., 2020). We have shown in our recent study that a delicate interplay of all three strategies is required to adequately control whole-body angular momenta in the sagittal and frontal planes (Matjačić et al., 2020). Control of COP is predominantly under the control of ankle muscles, which are a resource missing in the UTA population, while the control of lateral GRF is chiefly under the control of hip muscles, which are still at disposal after below-knee amputation. Thus, it is currently not known how UTA subjects would respond to a situation when an outward-directed perturbation was applied while being with the amputated leg in stance during slow walking.

The aim of this study was to investigate the kinematics and kinetics of dynamic balancing responses that UTA subjects use after outward-directed perturbing pushes to the pelvis while walking slowly on a treadmill. Our hypothesis was that the absence of ankle efferent and afferent function in UTA subjects, limiting their abilities to properly modulate COP and GRF under the prosthesis when perturbation commences while the amputated side is in-stance, would make use of “in-stance” balancing strategy impossible. Instead, they would be forced to resort to the “stepping” strategy commencing only with the ensuing step of the non-amputated limb. On the other hand, when UTA subjects face an outward-directed perturbation while in-stance on the non-amputated side it was our further hypothesis that their dynamic balancing responses would resemble those of the control population.

Methods

Participants

Fourteen UTA subjects (11 females, 3 males; age: 49.9 ± 12.4 years; body mass: 81.9 ± 15.2 kg; height: 173.7 ± 9.2 cm; left amputations: 9, right amputations: 5) and nine control subjects (2 females, 7 males; age: 46.7 ± 11.9 years, body mass: 79 ± 15 kg, height: 177.3 ± 7.4 cm) without known neurological and muscular-skeletal problems participated in this study. The inclusion criteria for UTA subjects were: unilateral transtibial amputation, at least 10 years of experience using a passive transtibial prosthesis, have had the current prosthesis for at least one year, the same medical doctor and prosthetist checked for socket fitness and alignment, have no problems with the prosthesis, k-levels K3 and K4 (Gailey et al. 2002), have no other muscular-skeletal problems, are able to walk independently without walking aids, and are able to follow instructions. The study was approved by the local ethics committee (number 45/2018) and all participants provided written informed consent.

Perturbing apparatus

The Balance Assessment Robot for Treadmill walking (BART), consisting of a wide instrumented treadmill and an actuated pelvic link with pelvis brace, was used to deliver perturbing force impulses at the level of the pelvis during walking on the treadmill. The pelvic link interacts with the participant’s pelvis within 5 actuated degrees of freedom (DoF) – translation in the sagittal, frontal and vertical directions,

pelvic rotation and pelvic list; the remaining DoF – pelvic tilt – is passive. Haptic interaction between the actuated pelvic link and the participant's pelvis is implemented via an admittance controller and can be arbitrarily set from very compliant (transparent mode – the interaction force between perturbations is as low as possible to allow free pelvis movement) to very stiff. During perturbation the period admittance controller delivers a force impulse to the pelvis. The duration and amplitude of the force impulse can be arbitrarily set within the range [0-1000 ms] and [0-200 N] respectively. Pelvis movement in the BART is measured with the pelvic link and is used to estimate the movement of center of mass (COM) in a similar way as reported in previous studies (Matjačić et al., 2017; Matjačić et al., 2018). GRF and COP signals are obtained by four precise force transducers (K3D120, ME Systeme GmbH) that are positioned underneath the treadmill. Left and right foot strikes are identified from instantaneous COP signals by a custom developed algorithm. The onset of perturbation is referenced with respect to foot strikes and can be arbitrarily delayed to a later time instance. Finally, the treadmill speed of walking can be arbitrarily set in a range up to 2 m/s. Detailed description of the BART is provided in previous studies (Olenšek et al., 2016; Matjačić et al., 2017).

Reactive Balance Assessment Protocol

Subjects started with a five-minute introductory session in order to familiarize themselves with the experimental conditions. They were secured with the pelvic brace, which in case of complete loss of balance would hold the subject erect and immediately stop the treadmill. They were instructed to walk within the central area of the treadmill – current pelvic position and the central treadmill area were visualized on a screen in front of the subject for orientation. Treadmill speed was set to 0.5 m/s and perturbation amplitude was normalized to 10% of each participant's body weight. The perturbation duration was set to 150 ms as in our previous studies (Matjačić et al., 2017; Matjačić et al., 2018). When no perturbing force impulses were delivered during the experiment the pelvic link of the BART was in transparent mode. At the beginning of the experiment each participant walked for approximately three minutes with no perturbations being delivered to assess their unperturbed gait. This initial period was followed by approximately 7 minutes of perturbed walking. During the perturbed walking force impulses were triggered at either the left or right foot contact in one of four perturbation directions to prevent predictability: forward, backward, inward or outward. Fig. 1 illustrates a top view of the experimental setup with indicated perturbation directions. All test subjects were subjected to seven repetitions for each perturbation direction on sides of the body. The time between two consecutive perturbations was randomly chosen with an interval of between 6 and 9 s. Altogether each subject received 56 perturbations that were block-randomized. With respect to our hypotheses only dynamic balancing responses following outward-directed perturbed walking were considered in further analysis.

Measurements and Data Analysis

The COM, COP and GRF signals were first segmented into strides. Gait cycle was defined as the period between two consecutive foot strikes of the same leg. Gait cycles immediately after the onset of outward-directed perturbation and gait cycles where no perturbation occurred were considered in the analysis. The

data of the selected gait cycles were segmented into the "in-stance" periods (from right foot strike to the next left foot strike or from left foot strike to the next right foot strike – from 0% to approx. 60% of gait cycle) and into the "stepping" periods (from left foot strike to the next right foot strike or from right foot strike to the next left foot strike – from approx. 60 to 100% of gait cycle) (Zadravec et al. 2020). In both periods and separately for unperturbed and outward-directed perturbed walking they were normalized to the duration of each period to allow visual comparison between different sub-phases of the in-stance and stepping periods of the gait cycles.

Data were organized into three groups: i) control group (responses to perturbations delivered at left foot strike in the group of control subjects; as in our previous study (Matjačić et al., 2019) we found responses to outward-directed perturbations for left and right side to be comparable), ii) non-amputated group (responses to perturbations delivered at the foot strike of the non-amputated leg in the group of UTA subjects) and iii) amputated group (responses to perturbations delivered at the foot strike of the amputated leg in the group of UTA subjects). For each subject COM, COP, GRF and durations of in-stance and stepping periods ($T_{in\text{-}stance}$ and $T_{stepping}$ respectively) were averaged across seven repetitions for unperturbed and outward-directed perturbed walking. If any of the seven repetitions markedly differed it was excluded from further analysis. At least five repetitions were averaged for each experimental condition. Peak excursions were obtained from averaged COM and COP for each subject and for outward-directed perturbed walking as well as for unperturbed walking. ΔCOM was calculated as the difference between COM peak excursion of outward-directed perturbed walking and COM peak excursion of unperturbed walking (sagittal plane – ΔCOM_{AP}; frontal plane – ΔCOM_{ML}). Likewise, ΔCOP was calculated as the difference between COP peak excursion of outward-directed perturbed walking and COP peak excursion of unperturbed walking (frontal plane – ΔCOP_{ML}). ΔCOM was determined once in the gait cycle while ΔCOP was determined separately for the in-stance and stepping period of gait cycles. Similarly, for outward-directed perturbed walking we calculated the deviation of durations of in-stance and stepping periods ($\Delta T_{in\text{-}stance}$ and $\Delta T_{stepping}$ respectively) from unperturbed walking.

We have further calculated the time integrals of outward-directed perturbed and unperturbed GRF for the in-stance and stepping periods. Finally, these integrals were normalized to the body mass of each subject and subtracted to yield force impulses for both planes (frontal plane – ΔGRF_{ML}) which acted against perturbation in both periods of the gait cycle. ΔCOM can be viewed as a controlled variable while ΔGRF and to some extent also ΔCOP (when the human body can be considered as an inverted pendulum) can be viewed as control variables. Thus, separately determining ΔGRF and ΔCOP for the in-stance and stepping periods provides information on the relative share of dynamic balancing responses coming from the in-stance strategy and stepping strategy.

Statistical Analysis

The normal distribution of data was tested using a Kolmogorov-Smirnov test. One-way analysis of variance (ANOVA) was conducted to compare the effect of group factor on $T_{in\text{-}stance}$ and $T_{stepping}$ of unperturbed walking in control, non-amputated and amputated groups. Likewise, for outward-directed

perturbation one-way ANOVA was conducted to compare the effect of group factor on $\Delta T_{in\text{-}stance}$ and $\Delta T_{stepping}$ in control, non-amputated and amputated groups. Finally, for each combination of outward-directed perturbation and in-stance or stepping periods one-way ANOVA was conducted to compare the effect of group factor (control, non-amputated and amputated) on ΔCOM, ΔCOP and ΔGRF. The Bonferroni method was used in post-hoc comparisons. The level of statistical significance was set to 5%. Data processing and data analysis were performed in MATLAB R2018b (The MathWorks, Inc.).

Results

Kinematics and kinetics of perturbed walking

Figure 2 shows kinematic and kinetic responses for a representative subject in each of the three groups during unperturbed and perturbed walking. The responses to outward-directed perturbations for a subject from the non-amputated group were similar to those for a subject from the control group. The response was characterized with COP_{AP} displacement under the stance leg toward the toes and with an increase in posterior GRF_{AP} that temporarily decelerated COM_{AP} (braking strategy). COP_{ML} was displaced toward the outer edge of the foot (medio-lateral ankle strategy) whereas an impulse-like increase in lateral GRF_{ML} acted to decelerate COP_{ML} movement (inertial strategy). Perturbation was contained completely by the described in-stance strategy.

On the other hand, in the responses of a subject from the amputated group we noticed a complete lack of in-stance response after outward-directed perturbation. Here the main strategy was first to move the non-amputated leg substantially more laterally to produce a “cross-step”. Once the non-amputated leg entered the stance phase of a gait cycle, which was substantially more laterally compared to unperturbed walking, the established lateral position of COP_{ML} in relation to COM_{ML} modified lateral GRF_{ML} which acted to decelerate COM_{ML} movement. This stepping response resulted in a substantially higher maximal excursion of COM_{ML} in the frontal plane.

Temporal parameters of unperturbed walking

Figure 3 shows durations of in-stance ($T_{in\text{-}stance}$) and stepping ($T_{stepping}$) periods for all three groups for unperturbed walking. Results show that the control group shared the duration of gait cycles approximately equally between in-stance and stepping periods. On the other hand, UTA subjects shortened the duration of gait cycle (combined $T_{in\text{-}stance}$ and $T_{stepping}$) with respect to control subjects and preferred to extend the duration of instance period $T_{in\text{-}stance}$ and shorten the duration of stepping period $T_{stepping}$ on the non-amputated side. Further statistical analysis showed a statistically significant effect of group factor on $T_{in\text{-}stance}$ ($F(2,34) = 21.27$; $p < 0.001$) for the three groups and post-hoc analysis showed statistically significant differences in all pair-wise comparisons between groups. The group factor had statistically significant effect also on $T_{stepping}$ ($F(2,34) = 12.4$; $p < 0.001$), however only

$T_{stepping}$ of the control group with respect to non-amputated and amputated groups was statistically different in the post-hoc comparison.

Temporal parameters of perturbed walking

Figure 4 shows changes in temporal parameters for the three selected groups after outward-directed perturbation. All groups have shown increased duration of in-stance period with respect to unperturbed walking. There was a statistically significant effect of group factor on $\Delta T_{in\text{-}stance}$ (F (2,34) = 16.4577; $p < 0.001$) and post-hoc comparisons showed that $\Delta T_{in\text{-}stance}$ for the amputated group was significantly smaller than $\Delta T_{in\text{-}stance}$ for the control and non-amputated groups. On the other hand, after perturbation all groups shortened the duration of stepping period $T_{stepping}$ with respect to unperturbed walking. Group factor did not have significant effect on $\Delta T_{stepping}$ (F (2,34) = 1.0512; p = 0.1143).

ΔCOM of perturbed walking

Figure 5 shows ΔCOM_{AP} and ΔCOM_{ML} for the three selected groups. ΔCOM_{AP} shows a substantial posterior shift of pelvis after outward-directed perturbation for control and non-amputated groups that was more pronounced in the control group. One-way ANOVA showed a significant effect of group factor on ΔCOM_{AP} (F (2,34) = 3.0192; $p < 0.001$) but no significant differences were found in post-hoc analysis.

In the mediolateral direction ΔCOM_{ML} indicates considerable and almost equal shift of pelvis in the direction of the outward-directed perturbation for the control and non-amputated groups. Even larger – almost double – pelvis shift in the direction of perturbation occurred in the amputated group. Statistical analysis showed that group factor had significant effect on ΔCOM_{ML} (F (2,34) = 38.5987; $p < 0.001$). Subsequent post-hoc comparison showed that ΔCOM_{ML} for the amputated group was significantly higher than ΔCOM_{ML} for the control group as well as for the non-amputated group.

ΔCOP and ΔGRF of perturbed walking

Figure 6 shows ΔCOP and ΔGRF for the three selected groups. In the in-stance period ΔCOP_{ML} shows a moderate shift of COP in the outward direction for control and non-amputated groups when compared to unperturbed walking whereas for the amputated group the shift was marginal. Statistical analysis showed a significant effect of group factor on ΔCOP_{ML} in the in-stance period (F(2,34) = 40.1967; $p < 0.001$) and post-hoc comparison showed that in the in-stance period ΔCOP_{ML} for the amputated group was significantly smaller than ΔCOP_{ML} for control and non-amputated groups.

In addition, ΔGRF_{ML} shows that GRF_{ML} in the in-stance period for the control and non-amputated group increased on average after perturbation when compared to unperturbed walking and decreased for the amputated group. Statistical analysis showed a significant effect of the group factor on ΔGRF_{ML} in the in-stance period (F(2,34) = 42.2117; $p < 0.001$) and further post-hoc analysis showed that in the in-stance

period ΔGRF_{ML} for the amputated group was significantly different than ΔGRF_{ML} for the control and non-amputated group.

In the stepping period ΔCOP_{ML} again shows a moderate shift of COP in the direction of perturbation for control and non-amputated groups when compared to unperturbed walking whereas the shift is substantial for the amputated group. Statistical analysis showed a statistically significant effect of group factor on ΔCOP_{ML} in the stepping period ($F(2,34) = 45.3744$; $p < 0.001$) and further post-hoc comparison again showed that in the stepping period ΔCOP_{ML} for the amputated group was significantly larger than ΔCOP_{ML} for control and non-amputated groups.

Finally, ΔGRF_{ML} also shows that GRF_{ML} in the stepping period for the control and non-amputated group slightly increased on average after perturbation when compared to unperturbed walking and increased substantially for the amputated group. Statistical analysis showed that the effect of group factor on ΔGRF_{ML} in the stepping period was statistically significant ($F(2,34) = 31.1269$; $p < 0.001$). Post-hoc analysis again showed that in the stepping period ΔGRF_{ML} for the amputated group was statistically larger than ΔGRF_{ML} for control and non-amputated groups.

Discussion

Dynamic balancing responses

In this study we investigated dynamic balancing responses following perturbations applied to the pelvis in the outward direction during slow treadmill walking in a group of control subjects and a group of high-functioning UTA subjects. Results show that the control group utilized the in-stance balancing strategy, which is consistent with the findings of Matjačić et al. (2020). Similarly, when UTA subjects were subjected to outward-directed perturbation upon entering stance phase with their non-amputated side they were also able to modulate COP and GRF in a similar way which indicates they used the in-stance strategy. However, when perturbations occurred when entering the stance phase on the amputated side UTA subjects did not show such COP and GRF modulation under their prosthetic leg which indicates a lack of in-stance balancing strategy. Instead, the balancing response commenced only after the non-amputated leg entered the ensuing stance when it was placed to make a cross-step. This indicates that their response mainly consisted of the stepping strategy. Delaying the corrective action resulted in higher COM displacement, which is in agreement with findings from other perturbation studies conducted in subjects without amputation and post-stroke subjects (Haarman et al., 2017; Matjačić et al., 2018, Zadravec et al. 2020). These findings support our hypotheses.

We found only one study that investigated dynamic balancing responses following unexpected perturbing pushes in the frontal plane applied to the pelvis of UTA subjects (Major et al., 2018). In that study subjects walked at a speed of 0.8 m/s and developed stepping responses regardless whether the non-amputated or amputated leg was in stance phase at the time of perturbation. These findings seem to contradict the results of the present study since in our study UTA subjects applied in-stance and not

stepping balance strategies when they counteracted outward-directed perturbation on the non-amputated side. Discrepancy between the two studies is to be attributed to different walking speeds which was in our study set to 0.5 m/s. Namely a recent study showed that balancing responses, when evoked by force impulses to the pelvis of subjects without amputation at walking speeds of 0.8 m/s and higher, predominantly consist of stepping responses while the contribution from in-stance responses is small. However, when the speed of walking is low (0.4 to 0.6 m/s) in-stance strategies are the primary mechanisms of dynamic responses (Matjačić et al., 2019).

Although the function of calf muscles was not directly assessed the outcomes of this study imply how important calf muscles are for efficiently counteracting perturbations directed in the outward direction during slow walking. It is interesting to observe that even though UTA subjects have in general well-preserved hip abductor muscles (Molina-Rueda et al., 2014), which have been shown to be the primary contributors to in-stance balancing response following outward-directed perturbation (Matjačić et al. 2018), there was a complete lack of modulation of lateral GRF on the amputated side during the stance phase. In our previous study we have shown that inertial strategy (modulation of lateral component of GRF) and braking strategy (modulation of posterior component of GRF) are closely coupled when subjects without amputation counteract an outward-directed perturbation that is delivered in double stance while walking very slowly (Matjačić et al., 2020). It has also been shown that this coupling is necessary to efficiently control angular momenta in the frontal and sagittal planes. However, since passive transtibial prostheses do not enable users to efficiently modulate COP, braking of the movement in the plane of progression, which is normally in the domain of calf muscles (Honeine et al. 2014), is not possible. Similarly, lack of braking was reported in a study where whole-body angular momentum fluctuations were investigated in amputee and non-amputee subjects across different walking speeds (Silverman et al. 2011). The authors suggested that reduced braking on the amputated side was associated with a significantly larger range of angular momentum in the sagittal plane in the first half of the gait cycle and that the absence of braking was a compensatory mechanism that helped to control angular momentum in the sagittal plane. Results of our study also show a lack of lateral COP displacement in the frontal plane, which has been shown in several previous studies to be essential in counteracting perturbation in the frontal plane (Hof et al., 2010; Hof and Duysens, 2018, Matjačić et al. 2019).

Our study also shows that even though the treadmill speed was set to be equal for all subjects, stride time during unperturbed walking was significantly shorter for UTA subjects than in control subjects. This implies that cadence was higher and steps were for UTA subjects shorter. They also spent considerably more time in stance on the non-amputated side than on the amputated side. This is in agreement with findings of other studies (Hak et al., 2013, Breuskens et al., 2014; Sheehan et al., 2015; Major et al., 2018) and may be related to the need of being capable to react faster to prospective perturbation with ensuing step. Since the perturbations in our study occurred at the beginning of the stance phase, adopting shorter and faster steps enabled the group of UTA subjects to react faster with the ensuing step of the non-amputated leg.

Clinical relevance

The perturbing paradigm used in this study bears close similarity with other perturbing modalities that one may come across during the activities of daily living. Perturbing pushes to the pelvis of a walking subject in the outward direction mimic situations where persons collide with other persons in a crowd. In this context results of this study suggest that inferior performance in counteracting a perturbing push when applied at the beginning of stance phase on the amputated side may also be an indication of inferior responses to collisions in everyday walking. Furthermore, significant displacement of COM from the base of support, resulting from sensory fusion errors in the internal balance model, without in-stance mechanisms to mitigate the effect of such gait instability inherently increases the risk of falling. To improve the dynamic balance in such situations rehabilitation after lower limb amputation should also incorporate task-oriented perturbation training as part of the rehabilitation program. Similar studies have shown that such training is an effective rehabilitation method to reduce falls in persons with lower limb transtibial amputation (Kaufman et al. 2014).

Development of active prostheses focused mainly on the aspects of improving propulsion on the impaired limb but did not explicitly consider dynamic balancing aspects, particularly not in the context of coping with unexpected perturbations (Windrich et al., 2016). Few studies investigated the effects of stiffness properties of the prosthetic leg on balance in the sagittal (Major et al., 2016) and frontal plane (Kim and Collins, 2017). They showed that appropriate control of stiffness in the artificial ankle joint may improve balance in terms of reduced variability and asymmetry in spatio-temporal parameters during unperturbed walking at self-selected speed. As an alternative, recent advances in the development of transtibial prostheses have produced a neuromuscular model-based control of prostheses (Thatte and Geyer, 2016; Markowitz et al., 2011). Such prostheses use neuromuscular models to determine the characteristics of actuators based on force, velocity and force feedback signals of the muscles. Studies show that such control of prostheses provide adaptive behavior at different walking speeds (Markowitz et al., 2011) as well as on different terrains, and can reproduce a normal walking pattern and successfully respond to trip-mimicking disturbances in the swing phase (Thatte and Geyer, 2016). Our study suggests that the developers and producers of active transtibial prostheses should consider incorporating functionality in their designs that could efficiently address the need of appropriate control of COP following an outward-directed perturbation. This could enable UTA subjects to more effectively cope with unexpected loss of balance during slow walking.

Methodological considerations and limitations of the study

In this study a relatively low walking speed of 0.5 m/s was selected for assessment of dynamic balancing responses. The rationale for selecting such a low speed of walking was to create conditions in which subjects typically apply in-stance balance strategies when subjected to outward-directed perturbations. A recent study with subjects without amputation shows that when walking on a treadmill with walking speeds ranging from 0.4 to 0.6 m/s, in-stance balancing strategies play the dominant role when counteracting perturbations. Exceeding this range of walking speeds first displays mixing of in-

stance and stepping strategies whereas at 0.8 m/s and above only the stepping strategy is used (Matjačić et al., 2019). Another aspect related to the choice of suitable walking speed is that starting and stopping of walking as well as changing directions are all maneuvers that are typically performed at reduced walking speed (Weerdesteyn et al., 2008).

We also focused on balancing responses to only one perturbation amplitude, i.e., 10% of body weight. Previous studies (Hof et al., 2010; Matjačić et al., 2019) have shown that such perturbation amplitude is strong enough to elicit substantial imbalance during walking without exciting undesirable leg pivoting or arm and trunk movement. Similarly, setting the onset of perturbation to the beginning of stance phase was motivated by the observation that it elicits the use of the in-stance strategy to the largest extent (Matjačić et al., 2019). It is our opinion that the selected experimental parameters (speed of walking, perturbation amplitude and onset of perturbation) determined experimental conditions in a way that challenged control and UTA subjects to the extent that allowed us to competently test the posed hypotheses.

The BART was controlled such that the interaction forces between the walking subject and the pelvis link were as low as possible. We have assessed the effect of interaction forces in a previous study and found that the influence of these forces on COP and GRF in the sagittal and frontal planes as well as on the EMGs of major lower limb muscles during unperturbed walking with walking speeds ranging from 0.4 to 0.8 m/s was negligible (Olenšek et al., 2017). In another study we have demonstrated that interaction between the balance assessment robot and the pelvis of a walking subject is purely passive, meaning that there is no exchange of energy between the walking subject and the BART except for the period when a perturbing push is delivered (Matjačić et al., 2017). Thus, we may consider that using the BART to deliver perturbations to the pelvis of walking subjects had negligible effects on the presented results.

Only high-functioning UTA subjects that were experienced and self-confident walkers have been included in this study. Although the results of this study cannot be directly applied to less experienced UTA subjects it is reasonable to assume that such situations demand greater effort that compels less experienced UTA subjects to rely on the non-amputated side even more. This needs to be further investigated.

Conclusion

The results of this study show how important the absence of ankle motor function is for efficiently negotiating outward-directed perturbations at slow walking speed. Since people after transtibial amputation lack this important resource, in-stance strategies do not constitute their dynamic balancing response after the perturbation occurring at the heel strike of the amputated side. Instead, their response begins with a substantial delay and concentrates on stepping strategy. Thus, the consequential stepping response is much less efficient compared to the in-stance response following a perturbation occurring at the heel strike on the non-amputated side which may be one of the potential causes of frequent falls among the UTA population.

The findings of this study suggest that to improve their balancing responses to unexpected balance perturbation people fitted with passive transtibial prostheses should undergo perturbation-based balance training during clinical rehabilitation. Likewise, the results of this study are relevant for the developers of active powered-prostheses, which should consider incorporating in their designs proper algorithms that would be able to identify a destabilizing event during walking and modulate COP and GRF accordingly under the prosthetic leg during stance phase.

Abbreviations

UTA – Unilateral Transtibial Amputation

COP – Center of Pressure

COM – Center of Mass

GRF – Ground Reaction Force

DoF – Degrees of Freedom

ΔCOM – Difference in peak values of COM between perturbed and unperturbed walking

ΔCOM_{AP} – Difference in peak values of anteroposterior component of COM direction between perturbed and unperturbed walking

ΔCOM_{ML} – Difference in peak values of mediolateral component of COM direction between perturbed and unperturbed walking

ΔCOP – Difference in peak values of COP between perturbed and unperturbed walking

ΔCOP_{AP} – Difference in peak values of anteroposterior component of COP between perturbed and unperturbed walking

ΔCOP_{ML} – Difference in peak values of mediolateral component of COP between perturbed and unperturbed walking

$T_{in\text{-}stance}$ – Duration of in-stance period.

$T_{stepping}$ – Duration of stepping period.

$\Delta T_{in\text{-}stance}$ – Difference in duration of in-stance period between perturbed and unperturbed walking

$\Delta T_{stepping}$ – Difference in duration of stepping period between perturbed and unperturbed walking

ΔGRF – Difference in GRF integrals between perturbed and unperturbed walking

ΔGRF_{AP} – Difference in anteroposterior component of GRF integrals between perturbed and unperturbed walking

ΔGRF_{ML} – Difference in mediolateral component of GRF integrals between perturbed and unperturbed walking

Declarations

Ethics approval and consent to participate

The study was approved by the Republic of Slovenia National Ethics Committee, application number 45/2018. All participants gave written informed consent.

Consent for application

Not applicable.

Availability of data and materials

The data used in this study may be available by the corresponding author upon a reasonable request to any qualified researcher.

Competing interests

The authors declare that they have no competing interests.

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Authors' contributions

MZ, HB and ZM contributed to the concept, research design and interpretation of the experimental results. HB recruited individuals with amputations that met inclusion criteria. MZ and AO performed perturbation-based measurements. MZ contributed to the signal processing, data analysis and prepared figures. ZM drafted the initial manuscript. AO wrote the final version of the manuscript. All authors critically revised the manuscript and approved the final version.

Availability of data and materials

The data used in this study may be available by the corresponding author upon a reasonable request to any qualified researcher.

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Figures

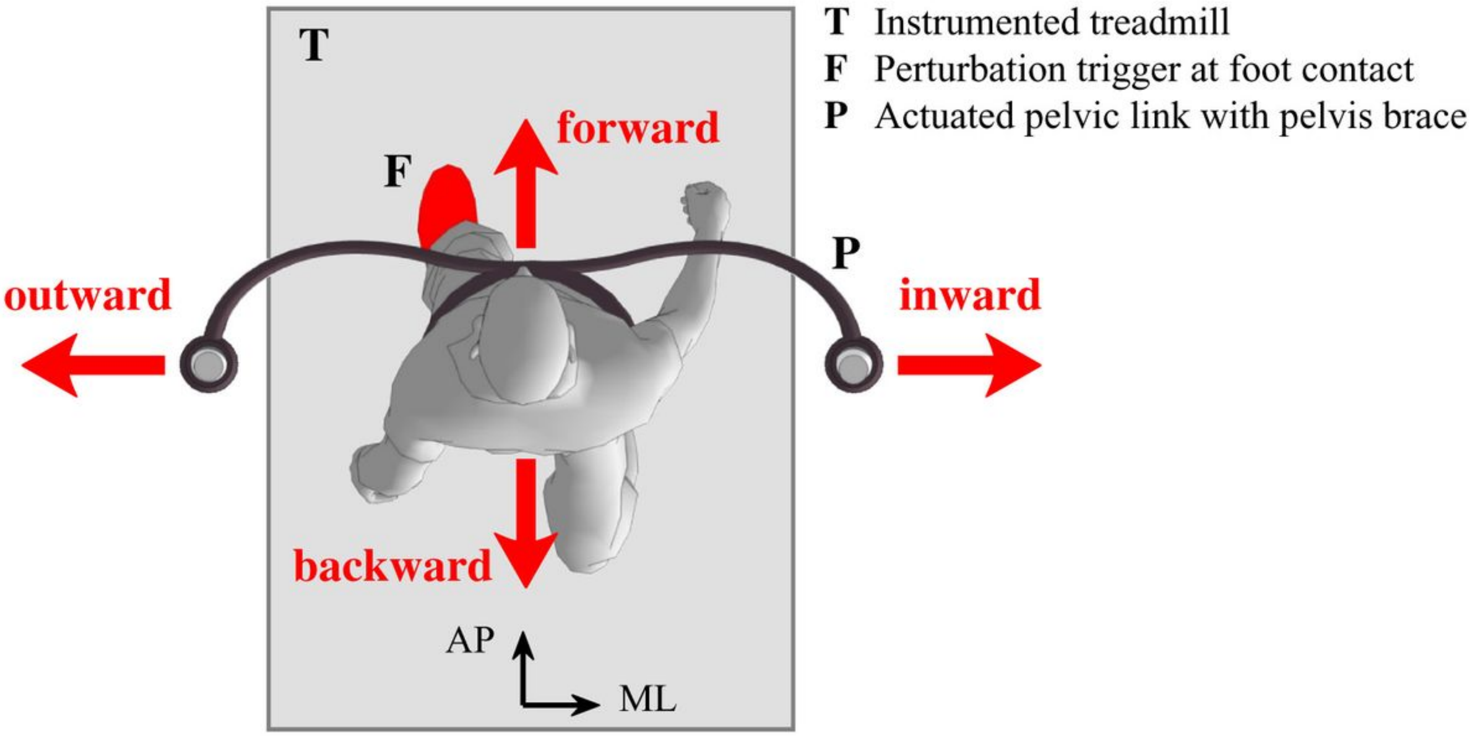

Figure 1

Experimental setup for assessing balance responses after perturbations applied to pelvis. Perturbations were applied in forward, backward, inward, outward directions and were triggered at either left or right foot strike.

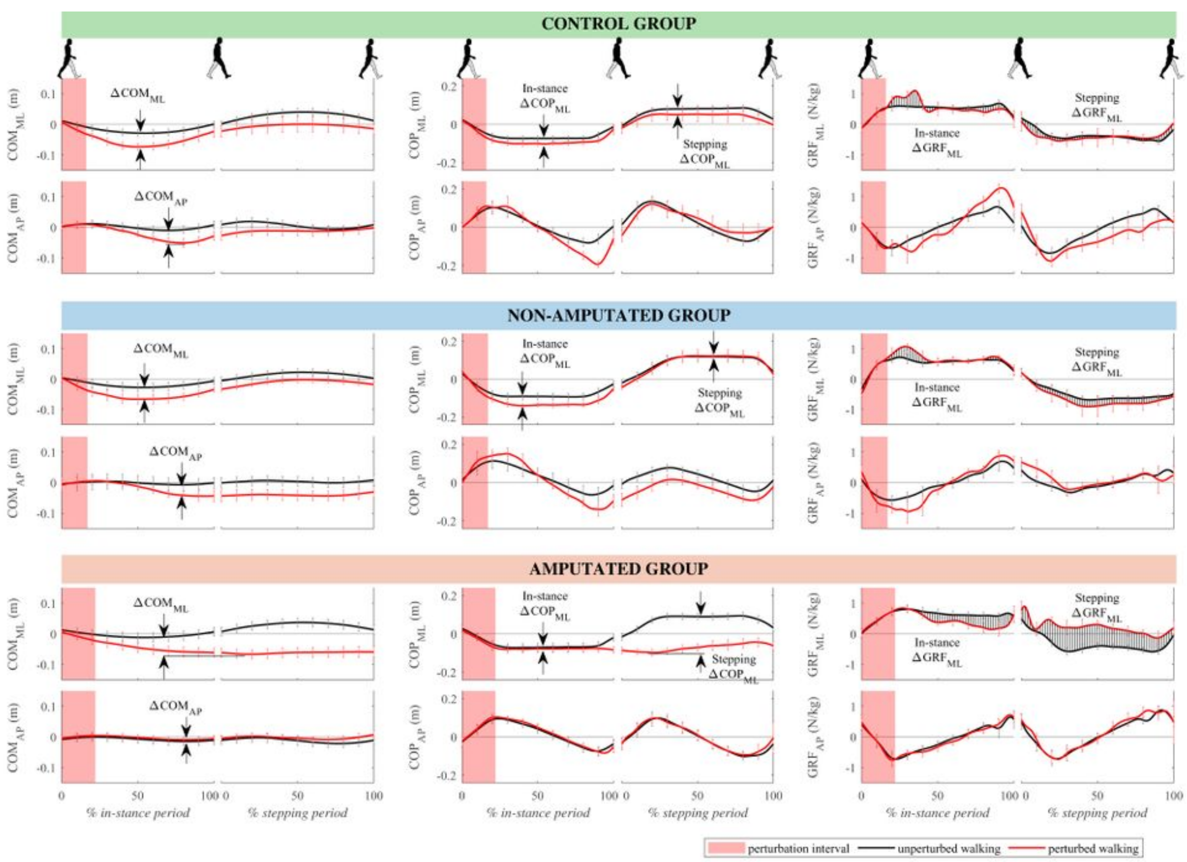

Figure 2

Kinematics and kinetics following outward-directed perturbation. Mean values and standard deviations are shown for the group representatives.

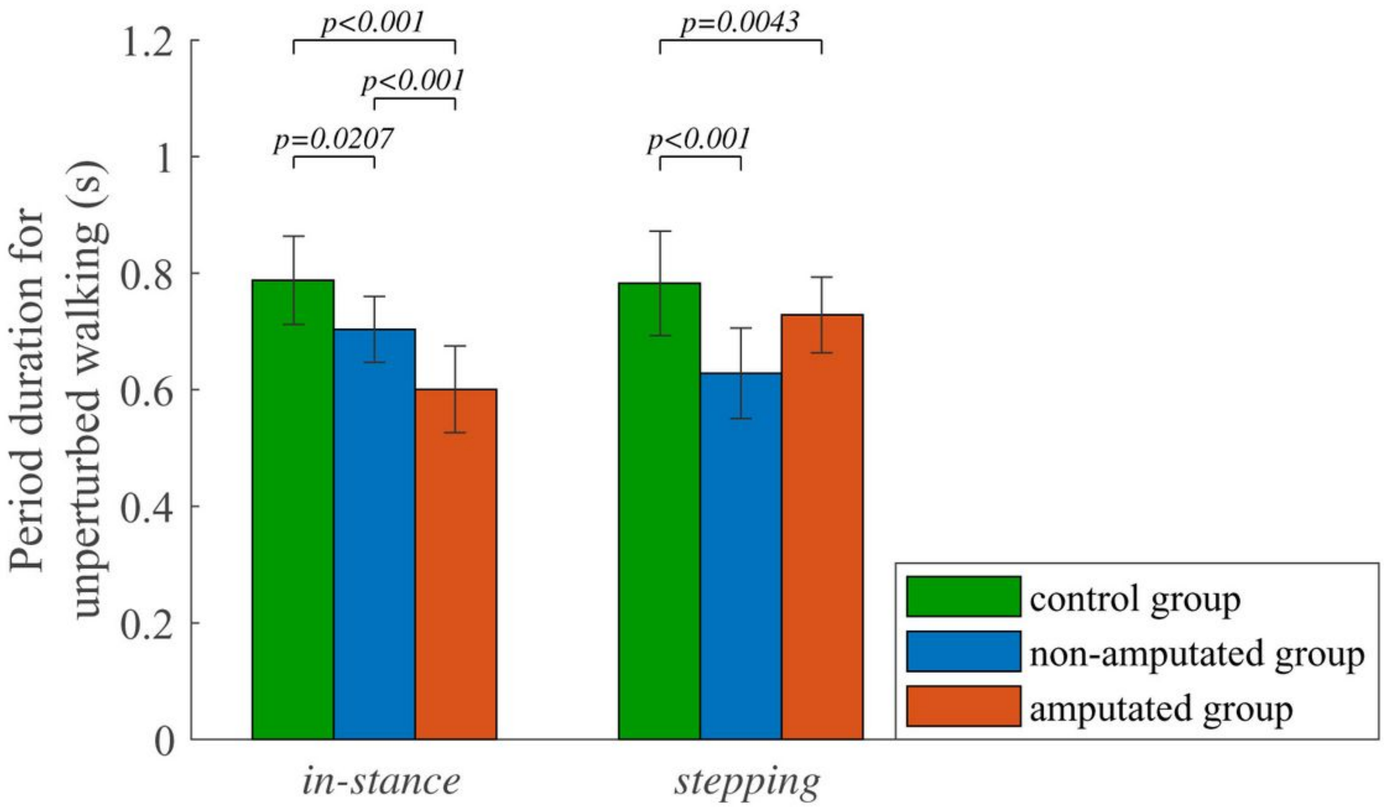

Figure 3

Group mean values and standard deviations for the duration of in-stance and stepping periods of unperturbed walking for all three groups. P-values indicate statistically significant effect of group factor and statistically significant differences between groups in Bonferroni post-hoc paired comparisons were found.

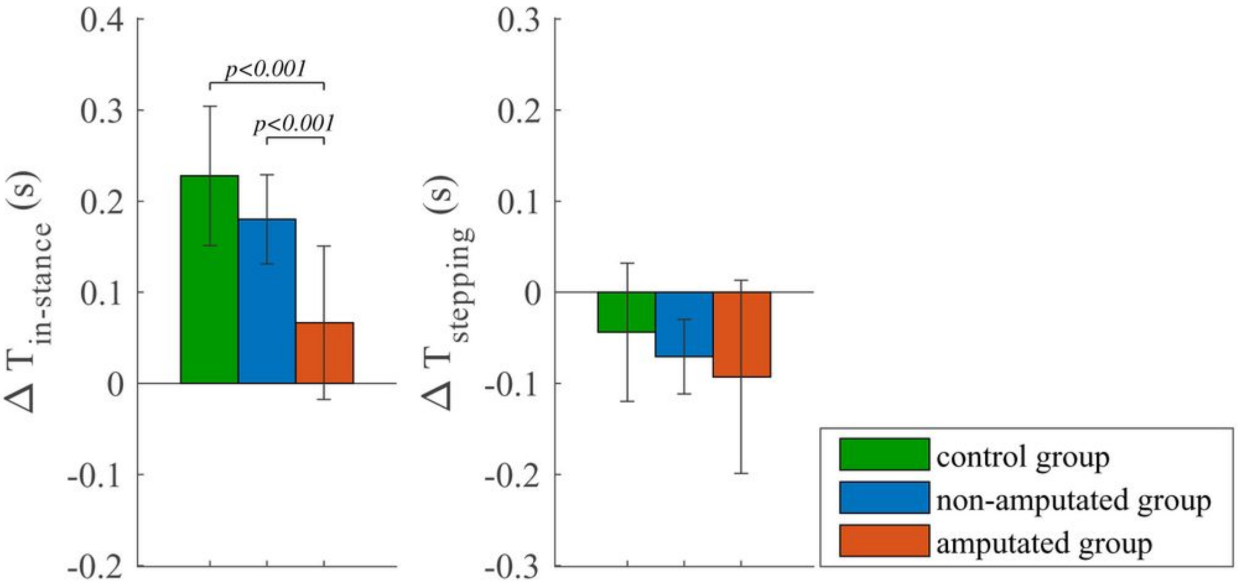

Figure 4

Temporal parameters of unperturbed and outward-directed perturbed walking. Group mean values and standard deviations are shown for all three groups. P-values indicate statistically significant effect of group factor and statistically significant differences between groups in Bonferroni post-hoc paired comparisons were found.

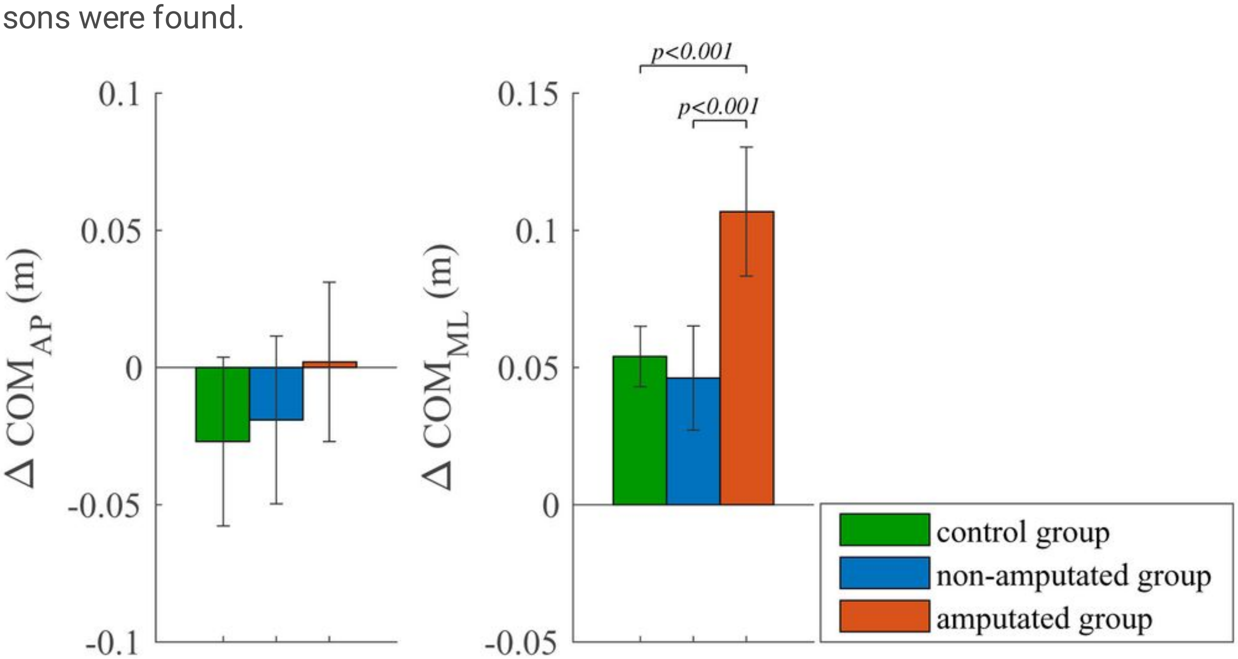

Figure 5

ΔCOM of outward-directed perturbed walking. Group mean values and standard deviations are shown for all three groups. P-values indicate statistically significant effect of group factor and statistically

significant differences between groups in Bonferroni post-hoc paired comparisons were found.

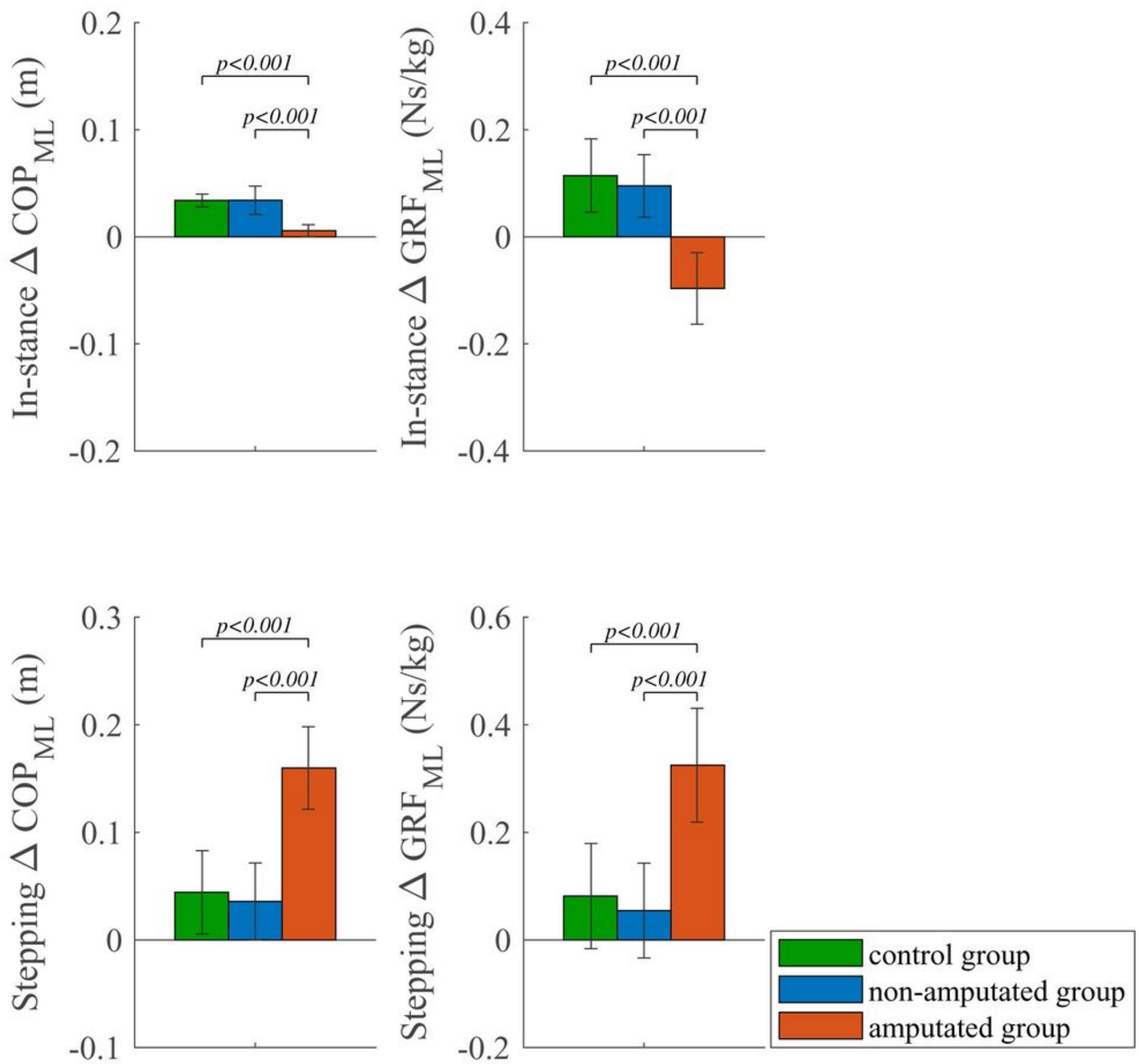

Figure 6

ΔCOP and ΔGRF of outward-directed perturbed walking. Group mean values and standard deviations are shown for all three groups. P-values indicate statistically significant effect of group factor and statistically significant differences between groups in Bonferroni post-hoc paired comparisons were found. Note that positive values of ΔCOPML indicate change in the direction of perturbation direction. Also note that positive values of ΔGRFML indicate change in the direction opposite to perturbation direction.