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Review

Postural Reactions to External Mediolateral Perturbations: A Review

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Abstract: Background: Mediolateral perturbations caused by external pulling and pushing forces can occur in everyday living. Although the loss of balance can appear in all directions, coping with frontal plane perturbations is more challenging. In literature, it is common to compare postural responses between the elderly and the young, but the accurate description of reactions in the young is lacking. This manuscript aims to review all previous papers investigating how healthy young adults deal with mediolateral perturbations in a standing position, including reviewing the methodology, outcomes, and sequence of responses in these studies. Methods: A systematic review was conducted of papers published from 1999 to 2022. The databases searched were PubMed, Google Scholar, ScienceDirect, and EBSCO. Eight publications met the inclusion criteria. Results: There is no consensus on the methodology for conducting this type of research and how to collect the data, as it varies between authors. Many papers lack an accurate description and justification of the magnitude of imposed perturbations. It has been shown that the first joint torque and muscle response on perturbation may not be produced by active and voluntary muscle contraction, but are due to tissue stiffness. Such stiffness-based balance control does not directly involve the CNS and provides an immediate and appropriate corrective response. Conclusions: Postural reactions to mediolateral perturbations are a complicated process that still need further, systematized investigation.

Keywords: mediolateral perturbations; pelvis perturbation; shoulder perturbation; young adults; balance; postural control

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1. Introduction

Maintaining a standing position is a task that requires proper neuromuscular coordination, the goal of which is to keep the center of mass (CoM) in the correct relationship with the base of support (BoS) [1,2]. The action of external forces is one of the factors that cause a disturbance in balance, forcing the body to take compensatory action to prevent a fall. Depending on the severity of the perturbation, the body must apply an appropriate strategy, and thus the sequence of activation of the muscles responsible for restoring balance must change. It is possible to regain stability by (1) adjusting the CoM relative to the BoS or (2) adjusting the BoS to the CoM [3].

In the first case, recovery can be achieved in two ways, depending on the severity and direction of the perturbation. In response to small, low-frequency perturbations, the body behaves as an inverted pendulum, which is mostly controlled by the ankle joint muscles. In response to higher, high-frequency disturbance, the body operates like a double inverted pendulum, in which the corrective action is carried out by the hip and trunk muscles [4]. When perturbations are extensive, the projection of the CoM exceeds the BoS area. At this point, two other strategies are used to recover the balance. The first is the loaded side-step (LSS) strategy, which involves taking a side step with the limb initially passively loaded due to the perturbation. The second option is the unloaded crossover step (UCS) strategy, during which a side-step is performed with the limb primarily passively unloaded. In this strategy, the limb performing the step crosses with the supporting leg [5,6].

As mentioned by Welch and Ting [7], postural control is a fundamental motor task that must adapt quickly to a dynamically-changing environment. This is mainly noticeable when unexpected perturbations occur. According to Petró et al. [8], perturbations can be (1) mechanical stimuli (sudden perturbation like pulling, pushing, and tugging or continuous motorized movement of support surface); (2) sensory stimuli (visual, vestibular, and proprioceptive); or a (3) combination of such perturbations [8–10]. The direction of the application of the disturbance also seems to be an important factor. Williams et al. [11] emphasize that the response to mediolateral perturbation is more challenging as compared to anterior–posterior ones. It is worth noting that it is much easier to find papers in which the reactions of the body (both during gait and standing) [12] to perturbations taking place in the sagittal plane are analyzed. A considerably smaller number of authors undertake to evaluate the responses to perturbations occurring in the frontal plane. Since perturbations in the frontal plane induce an asymmetrical response, it has been observed that the postural control is dominated by the limb loading/unloading mechanism (aforementioned LSS and UCS strategy), which is predominantly controlled by the hip abductors and adductors [13–15]. It is worth mentioning that Nashner and co-workers [4,16,17] have emphasized that almost 95% of the anterior–posterior sway happens around the ankle and the hip axis, due to the reduction of the high number of degrees of freedom by compressing them into two muscular synergies termed the ankle and hip. Therefore, it seems interesting to see if the same mechanism occurs when the body responds to mediolateral perturbations that do not lead to a step strategy.

Therefore, the purpose of this paper was to review all previous papers investigating how healthy young adults deal with mediolateral perturbations in a standing position, including reviewing the methodology, outcomes, and sequence of responses in these studies. To this end, four sequences were established showing the responses, successively of the ankle, knee, pelvis, CoP, and CoM, as a function of time against destabilizing stimuli in the frontal plane. It will allow systematizing responses to the mediolateral equilibrium perturbations occurring in the daily environment.

2. Materials and Methods

2.1. Search Strategy

One author (RB) searched the four electronic databases PubMed, EBSCO, Google Scholar, and Science-Direct to find relevant papers. The databases were searched in July–August 2022. Only full-text articles in English, written since 1999, were included. Detailed search terms are in Table 1. The differences in the search terms used for the listed databases are due to the need to limit the search. The PubMed, Google Scholar, and EBSCO databases displayed too many records that did not coincide with the topic.

Table 1. The search term combinations applied within each of the four electronic databases.

PubMed	Google Scholar	ScienceDirect	EBSCO
(postural control) AND (perturbation) AND (medio-lateral)	(postural control) AND (perturbation) AND (medio-lateral)	(postural control) AND (perturbation) AND (medio-lateral)	(postural control) AND (perturbation) AND (medio-lateral) NOT (treadmill)
			(postural control) AND (medio lateral) AND (pushing) NOT (treadmill)
(postural control) AND (medial perturbation) NOT (treadmill)	(postural control) AND (medial perturbation) NOT (treadmill) NOT (walking) NOT (walk) NOT (older) NOT (disease) NOT (unilateral) NOT(children) NOT (monopedal) NOT (cortex)	(postural strategy) AND (balance) AND (medio lateral direction perturbation) AND (standing)	(postural control) AND (medial perturbation) NOT (treadmill)

Table 1. *Cont.*

PubMed	Google Scholar	ScienceDirect	EBSCO
(postural control) AND (medio lateral) AND (perturbation) AND (standing)			(postural control) AND (medio lateral) AND (perturbation) AND (standing)
(postural equilibrium) AND (medio lateral) AND (perturbation) NOT (gait)			(postural equilibrium) AND (medio lateral perturbation) NOT (gait)
(postural control) AND (standing) AND (external perturbations) AND (healthy adults)			(postural control strategy) AND (medio lateral) AND (perturbation)

2.2. Eligibility

The inclusion criteria were: (1) human participants, (2) participants of legal age but not elderly, (3) participants without disease, (4) bipedal starting position on a flat surface, (5) induction of perturbation by pushing or pulling at the level of the hips or upper limb girdle, (6) immobile ground, (7) availability in the English language.

2.3. Review Process

Duplicate articles from different sources were removed. The three-step process for qualifying search results was carried out by the authors (RB and MB). First, it was checked whether the title met the inclusion criteria. If so, the abstract was read to determine whether the relevant parameters were studied. If the abstract did not provide all the necessary information, it was searched in the full text. The reference lists of eligible articles were manually searched by one of the authors (RB) to find additional studies for this review. Checking this process was done by another author (MB). Papers investigating postural responses during movement, single-leg stance, on custom surfaces (foam), or those focusing on perturbations induced by the subject (such as voluntary elevation of the upper limb), were excluded. Publications in which research was conducted with the disabled, as well as children or the elderly, were also excluded. Studies involving elderly subjects were included in publications comparing responses between elderly subjects and young adults, as long as it was possible to obtain information on the postural responses in young subjects. One reviewer (MB) compiled all articles using a reference managing software (EndNote X7.7, Clarivate Analytics, Philadelphia, PA, USA).

Next, one of the authors (RB) analyzed the content of the selected publications in terms of five factors. These included: (1) characteristics of the study group, (2) the purpose of the study and the participants' task, (3) perturbation characteristics, (4) measured outcomes, and (5) the sequence of inclusion of kinematic and kinetic parameters in response to the perturbations. The characteristics of the study group involved the number of participants, their gender, mean age, body weight, and height. Characteristics of perturbations were formed by reviewing the location (waist or shoulder), method of force application (pushing or pulling), the manner of determining that force, and the number of perturbations. In addition, if possible, measurement equipment and data recording frequencies were determined. Information on the predictability of perturbations was also sought. Finally, the findings and results of each paper were studied and summarized in a table. Data on the sequence of changes in muscle activity (EMG and, in one case, MMG), ground reaction forces, joint angles, and torques were collected from the results sections of the analyzed papers and placed into four summary tables (Appendix A). The tables were for the ankle joint, knee joint, hip joint, and CoP with CoM. Events in each table were arranged chronologically. If required data did not appear directly in the text or the table, an attempt was made to estimate the numerical values from available graphs.

2.4. Quality Assessment

Each of the eligible papers was assessed for methodological quality using a checklist for randomized and non-randomized studies [18]. The checklist consisted five sub-scales: (1) reporting (10 items), where it checked whether the information contained in the article was sufficient for an unbiased evaluation of the study results; (2) external validity (three items), to determine the extent to which the retained results can be generalized to the entire population; (3) bias (seven items), to address bias in the measurement of intervention and outcome; (4) confounding (six items), to determine bias in the selection of study participants; (5) power (one item), where it was attempted to assess whether negative study results could be due to chance.

Due to the nature of included publications, not all areas of the checklist were suitable for quality assessment. Therefore, only certain parts of the checklist were selected for use. These were: all items from the reporting area, one item (item no. 11) from the external validity area, items no. 14, 15, and 18 from the bias area, the confounding area was excluded, and item no. 27 from the power area. The total score possible to be gained was 19 points.

3. Results

Initially, the process of screening the electronic database yielded 11739 papers. Screening of the titles and abstracts eliminated 11728 articles. The agreement was reached for 11 papers that were related to the purpose of this review. Further analysis of the study excluded another three full-text records due to the inability to obtain specific numerical data. A total of eight articles were selected for the review process (Figure 1).

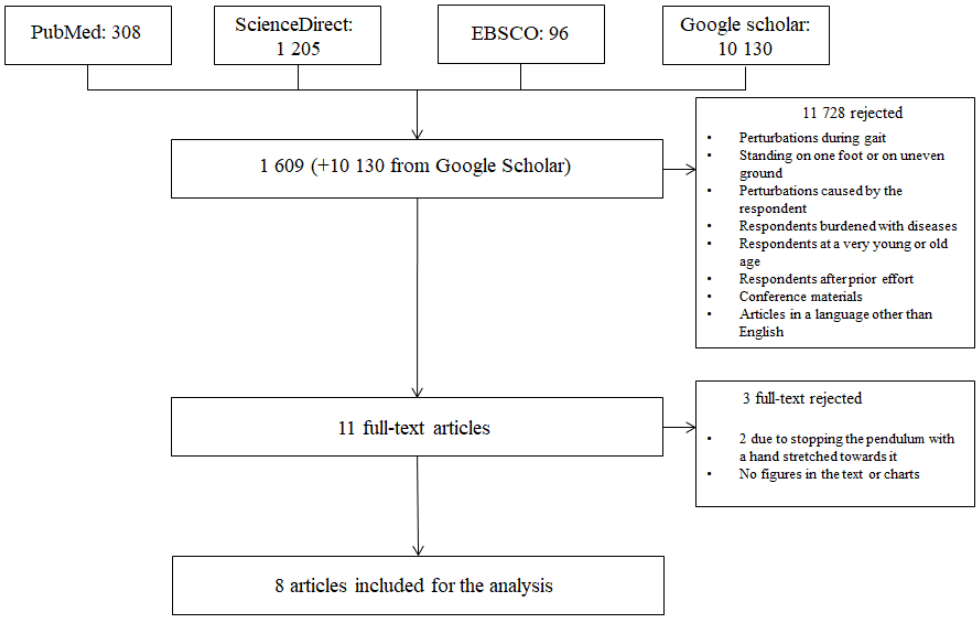


Figure 1. Flowchart demonstrating the selection of articles through the review process.

The following items were selected from the eight papers that were included in this review: characteristics of the people who participated in the study, the purpose of the study, and the tasks participants had to perform. The other features were a description of the equipment, perturbation locations, and the study results (Table 2). The papers were arranged in order from oldest [19] to newest [20].

Table 2. Data extracted from reviewed papers.

Study and Quality	Study Group: Age (Years); Body Mass (kg); Body Height (cm)	Aim/Partipant Task	Equipment/Perturbation Source/Place of Impact/Essential Description/Start of the Measurement	Results/Findings
Luchies et al. (1999) [19] Quality: 13/19	Young (Y): 13 W Age: 20.9 ± 2.5 Weight 62 ± 5.4 Height: 170.3 ± 6.9 Elderly (E): 11 W Age: 68 ± 4.1 Weight: 69 ± 14.6 Height: 165.8 ± 6.8	Aim: to investigate whether the performance on an involuntary step task was comparable to the performance on a voluntary step task. Subjects stepped as fast as possible in the direction of a minimally destabilizing lateral waist pull (voluntary step task), or they responded naturally to a large destabilizing lateral waist pull (involuntary step task).	Kinematics-Optotrak System (200 Hz). Two force plates measured the GRF from both legs (2000 Hz). EMG bilateral (2000 Hz): TA. Pulling/Pelvis. Safety harness. The second harness around the waist incorporates padded blocks. Lateral pulls were produced using electronically released weight and cable systems. The waist pull distances were 8.7% and 1.3% of the waist height. Start of the measurements: when the waist pull force exceeded a 5 N.	(1) In the voluntary step task, the elderly required significantly more time to raise their foot than the young (Y: 307 ± 50 ms; E: 425 ± 116 ms). (2) In involuntary tasks, both groups were equally fast in foot lifting (Y: 323 ± 59 ms; E: 334 ± 50 ms). (3) The voluntary step task underestimates the elderly ability to respond quickly when encountering large perturbations.
Rietdyk et al. (1999) [15] Quality: 13/19	10 M Age: 26 ± 4.2 Weight: 86 ± 8.2 Height: 181.6 ± 7.1	Aim: to provide insights into the underlying motor mechanisms used in postural control by determining the joint moments during balance recovery from ML perturbations. Participants were instructed that they would be receiving pushes to the trunk or pelvis, and that they should attempt to maintain balance without taking a step.	42 infrared emitting diodes (IREDs) were placed on the participant; three IREDs (40 Hz) were placed on the perturbation device; uni-axial force transducer. Two force plates (480 Hz). Pushing/Pelvis and Shoulder. The impulse of ML perturbation: Shoulder: 13.1 to 28.9 Ns; Pelvis: 16.8 to 31.4 Ns. Start of the measurements: Bias + 2SD.	(1) Muscle stiffness, not reflex-activated muscle activity is responsible for first joint torques. (2) The direction of trunk movement was dependent upon the location of perturbation. At the shoulder level, both the legs and trunk were displaced in the same direction as the perturbation. (3) In response to the pelvis perturbation, the legs moved in the same direction as the perturbation, while the trunk moved in the opposite direction.

Table 2. *Cont.*

Study and Quality	Study Group: Age (Years); Body Mass (kg); Body Height (cm)	Aim/Partipant Task	Equipment/Perturbation Source/Place of Impact/Essential Description/Start of the Measurement	Results/Findings
Matjacić et al. (2001) [21] Quality: 13/19	8 M Age: 31 ± 6.7 Weight 75.8 ± 11.4 Height: 182 ± 8.8	Aim: to assess functional postural responses by analyzing the net joint torques (NJT) in the ankles and the hips, resulting from perturbations delivered in multiple directions to subjects standing quietly. To stand relaxed prior to perturbation and attain the same posture throughout the trial after recovering from perturbation.	Two AMTI force plates (100 Hz). Multi-purpose rehabilitation frame (MRF) consists of two 2-dof rot joints, two 1-dof rot joints, two vertical supportive rods and a bracing system. Cylindrical pegs were used to constrain the position and orientation of the feet. Inclination angles of the MRF collected by potentiometers (Spectrol 157-9002-103) mounted to one of the 2-dof joints of the MRF. Pulling/Pelvis. Max perturbation amplitude: 34–60 Nm. Start of the measurements: 500 ms prior to the beginning of the perturbation and lasting for 3.5 s.	(1) The ankle sum NJT (in AP direction), and the ankle and hip sum NJT (ML) are the global variables being controlled. (2) The CNS controls the recovery from the multiple-direction perturbations of moderate strength, by decoupling the AP–ML postural space into two orthogonal directions (AP and ML).
Mille et al. (2005) [5] Quality: 13/19	Y: 10 Age: 24 ± 1.4 Weight: 60.9 ± 6.3 Height: 167.3 ± 5 E: 10 Age: 73.3 ± 6.3 Weight: 62.7 ± 14.2 Height: 159.3 ± 5.1	Aim: to determine age-related differences in protective stepping behavior in response to lateral waist-pull perturbations of postural balance. Subjects were instructed to react naturally to prevent themselves from falling.	Kinematics-Motus six-camera system (60 Hz). Two force plates (500 Hz). Pulling/Pelvis. Safety harness. Protective stepping responses were evoked by a motor-driven, waist-pull system [22]. The lateral pulls of constant magnitude (amplitude: 22.5 cm; velocity 31.5 cm/s; acceleration: 900 cm/s^2). Two cables were attached on one end to a rigid connection on a belt at the level of the subject's waist and on the other end to the puller system. Start of the measurements: 5 s after start of kinematics and GRF recording.	(1) The young used mainly LSS. The elderly preferred UCS and faced more inter-limb collisions. (2) When the elderly used LSS, steps were longer, slower, and higher. (3) The elderly generated more, slower hip adductor torque in single-leg support. (4) Impaired hip abduction torque-time capacity, lateral lumbar mobility, and trunk control may be responsible for the difficulty of lateral balance control in the elderly.

Table 2. *Cont.*

Study and Quality	Study Group: Age (Years); Body Mass (kg); Body Height (cm)	Aim/Partipant Task	Equipment/Perturbation Source/Place of Impact/Essential Description/Start of the Measurement	Results/Findings
Mille et al. (2013) [23] Quality: 12/19	Y: 26 (7 M, 19 W) Age: 23.5 ± 3.2 Weight: 70.3 ± 13.4 Height: 169.6 ± 9 E: 49 (11 M, 38 W) Non-faller: 30 (no data) Age: 72.5 ± 5.9 Weight: 66.4 ± 13.1 Height: 163.5 ± 8.2 Faller: 19 (no data) Age: 75.2 ± 7.8 Weight: 68.1 ± 12.1 Height: 161.2 ± 5.3	Aim: to determine the stepping response patterns evoked by different directions of externally-applied postural disturbances in younger and older adults, in relation to falls. To react naturally to prevent themselves from falling.	Kinematics-Motus six-camera system (60 Hz). Two force plates (500 Hz). Motorized pelvic-pull system. Balance perturbed through a pulley cable and switching system (displacement: 18 cm; velocity: 36 cm/s, acceleration: 720 cm/s^2). Pulling/Pelvis. Start of the measurements: 5 s after start of kinematics.	(1) The young used mainly LSS. (2) The elderly used mainly several steps, more often after ML perturbations. (3) The elderly faced inter-limbs collisions much more often. (4) With aging, balance recovery by stepping is harder for the lateral direction.
Lee et al. (2019) [24] Quality: 13/19	11 (5 M, 6 W) Age: 28.09 ± 4.35 Weight: 71.09 ± 18.75 Height: 167 ± 7	Aim: to examine the effects of external perturbation and landing orientation on ML control stability in step initiation. To stand still with feet shoulder-width apart, and even body weight distribution between feet.	One force platform AMTI (1000 Hz). EMG bilateral (Myopac, 1000 Hz): TA, MG, RF, BF, GMed, EO, RA, ES. Two accelerometers: (1) the pendulum accelerometer (model 208CO3, PCB Piezotronics Inc, USA); (2) the participant accelerometer (model 1356a16, PCB Piezotronics Inc, USA) on the dorsal surface at the level of L5/S1. A pendulum (30 cm long wooden stick) and a flag. Pushing/Shoulder. A load (5% of the individual's body mass) was attached to the pendulum. Initial angle of 30° to the vertical (0.8 m from the body). Released 1–2 s after the start of data collection. Start of the measurements: 0 ms indicates pendulum release.	(1) The temporal events of CoP and pelvic movement were not significantly different. (2) Most of the segments showed reciprocal muscle activation patterns in relation to the perturbation release time. Subsequently, all segments showed co-contraction muscle activation patterns, which were significantly affected by step side, perturbation, and orientation. (3) The results suggest that the way in which the CNS initiated a step was identical to the CoP, and then pelvic movement. (4) The outcome highlights the importance of external perturbation and foot landing orientation effects on postural adjustments, which may provide a different approach to help step initiation.

Table 2. *Cont.*

Study and Quality	Study Group: Age (Years); Body Mass (kg); Body Height (cm)	Aim/Partipant Task	Equipment/Perturbation Source/Place of Impact/Essential Description/Start of the Measurement	Results/Findings
Inacio et al. (2019) [25] Quality: 13/19	Y: 15 (7 M, 8 W) Age: 29.1 ± 1.1 Weight: 70 ± 3.6 Height: 174 ± 7 E: 15 (9 M, 6 W) Age: 71.3 ± 0.9 Weight: 80.4 ± 4.1 Height: 170 ± 2	Aim: this study investigated the influence of hip abductor–adductor neuromuscular performance on the weight transfer phase of lateral protective stepping. To stand still and recover balance with a single lateral step.	Kinematics-Vicon System (120 Hz). Two AMTI force plates (600 Hz). EMG bilateral (1500 Hz): GMed, TFL, ADD. A motorized waist-pull system [22]. To laterally shift their weight to distribute 50%, 65%, or 80% of BW onto the pre-determined stepping limb, using real-time visual feedback of the vertical GRF provided by a monitor. Pulling/Pelvis. Start of the measurements: moment of perturbation triggering.	(1) During the lateral balance perturbations, the elderly had a lower incidence of lateral steps, reduced hip abductor–adductor (AB–AD) neuromuscular activation (RActv), and delayed weight transfer. (2) The center of mass momentum at step onset, step-side peak rate of vertical force development, hip AB net joint torque, and power were larger in the elderly group. (3) The older adults had greater hip muscular output during the weight transfer phase, but their lateral balance recovery was still impaired. (4) The reduced maximal hip AB–AD capacity, especially RActv, may have been a greater contributor to this impairment, as it affects the ability to generate rapid force, crucial for balance recovery.

Table 2. *Cont.*

Study and Quality	Study Group: Age (Years); Body Mass (kg); Body Height (cm)	Aim/Partipant Task	Equipment/Perturbation Source/Place of Impact/Essential Description/Start of the Measurement	Results/Findings
Zhu et al. (2022) [20] Quality: 12/19	12 (6 M, 6 W) Age: 20.9 ± 0.7 Weight: 58.3 ± 6.2 Height: 169.9 ± 6.9	Aim: to investigate how rapidly the lower-limb muscles and joints would respond to the unexpected standing balance perturbations. To stand still and recover balance with a single lateral step.	Kinematics- eight-camera Vicon System (250 Hz). Two AMTI force plates (1000 Hz). EMG (2000 Hz) unilateral: TA, GMed, RF, ST, IL, GMax, AM. The waist-pull, hardness system. The maximal ML pulling displacements was set as 4%, 6%, and 8% of each participant's height. The pulling magnitude corresponded to the 1/3, 2/3, and 3/3 of the maximal pulling displacement, respectively. Each pull's duration, displacement, and velocity were measured based on the flash time of infrared light. Pulling/Pelvis. Start of the measurements: start of balance perturbation.	(1) Muscles responsible for resisting perturbations were activated earlier than their agonists. (2) In response to any perturbation, the ankle muscles had the fastest rate of activation. (3) Lower limb joint moments consistent with perturbation had a faster increase. (4) Muscle activity was faster with greater perturbations, but not joint kinetics and kinematics.

Abbreviations: M—men; W—women; GRF—ground reaction force; NJT—net joint torques; BW—body weight; ML—mediolateral; TA—tibialis anterior medial; MG—gastrocnemius; RF—rectus femoris; BF—biceps femoris; GMed—gluteus medius; GMax—gluteus maximus; EO—external oblique, RA—rectus abdominis, ES—erector spinae, TFL—tensor fasciae latae, AM—adductor magnus; ST—semitendinosus, IL—iliopsoas; ADD—adductor magnus; AB–AD—abductor–adductor; RActv—neuromuscular activation.

3.1. Characteristics of Participants

In all publications, the average age of participants ranged from 20 to 32 years. In two publications [15,21], the authors conducted the research with men only, and in one with women [19]. The remaining papers included participants of both sexes, although Mille et al. [5] did not report how many were women versus men.

The authors of papers [5,19,23,25] also included older people. The average age in this group ranged from 68 to75.2 years.

All authors included information on the average height and weight of the study participants. Mille et al. [23] provided the mean, median, maximum, and minimum values for height and weight. Inacio et al. [25] and Zhu et al. [20] also calculated the mean BMI values of study participants.

The most detailed characterization of the study group was provided by Zhu et al. [20]. In addition to the previously mentioned parameters (height, weight, BMI), the authors reported the participants' dominant limb and its length, the results of the IPAQ-S, and FES-I short version. The dominant limb was also taken into account by Lee et al. [24].

3.2. Ways and Levels of Inducing Mediolateral Perturbations

Lateral perturbations can be induced in two ways: by pulling or pushing, and at two levels: pelvic and shoulder. The first method (pulling) was the most common, as it was opted for in five publications [5,19,20,23,25]. The pulling mechanism usually consisted of an electronically controlled motor with cables. The cables were connected with a harness worn by subjects. Activation of the machine resulted in the generation of perturbing force. Zhu et al. [20] used their design of a motor-based pulling device. Inacio et al. [25], Mille et al. [5] and Mille et al. [23] used a pulling system designed by Pidcoe and Rogers [22], that involved a closed-loop stepper motor which induces perturbations. Luchies et al. [19] utilized a cable system connected to a certain weight. The weight release was operated electrically.

Two authors, Rietdyk et al. [15] and Lee et al. [24] decided to generate perturbations by pushes. Since Rietdyk et al. [15] preferred unpredictable perturbations, they chose the most simple method, i.e., pushing the participants with a stick. They believed that inducing perturbations with any mechanical device would make them easier for subjects to predict. In opposition were Lee et al. [24] who used a pendulum of a certain mass and a flag. The authors wanted the perturbations to be easily predictable.

Matjacić et al. [21] used a multi-purpose rehabilitation frame (MRF). The MRF was designed by Matjacić et al. [26]. It consisted of a frame with two two-degrees-of-freedom joints at the level of the subjects' ankles, and two one-degree-of-freedom joints at the subjects' hips level. The MRF was attached via a bracing system on the cross-bar. The movement was possible due to servo-controlled actuators.

The vast majority of authors, i.e., six [5,19–21,23,25], took the lower limb girdle as the place of perturbation application. Lee et al. [24] used a pendulum with a load that knocked the study group off balance by pushing at the shoulder rim, while Rietdyk et al. [15] studied pushes through both the upper and lower limb rims.

The method of determining the force applied to throw off balance was not always clearly defined. In three publications, it was 20% [19], 8% [20], or 5% [24] of the body weight. Additionally, Zhu et al. [20] applied forces of three magnitudes. The maximum force was 8% of the body mass, medium force 5.28% (0.66*8%) of the body mass, and minimal force 2.64% (0.33*8%) of the body mass. Matjacić et al. [21] conducted a pilot study. On this basis, they determined that suitable perturbation magnitude for the conducted study would be such when it was easy for the subjects to maintain balance after a backward pull. Similarly, Luchies et al. [19] selected the strength of the perturbation so that it was necessary for the subjects to take a step to regain balance. In three publications [5,15,23], explanation of chosen parameters was not given.

3.3. Measured Outcomes

Basis for all measurements is the defining moment when the recordings begin. In most cases, it was when perturbing force started affecting a subject. However, Matjacić et al. [21] and Lee et al. [24] decided to start measurements when the perturbation device started its motion, i.e., before any force started affecting a subject.

In all publications, the authors chose to measure ground reaction forces. In all but one paper [24], ground reaction forces were measured using two platforms. The recording frequency varied between 100 and 2000 Hz, and was different in each paper: 100 Hz [21], 480 Hz [15], 500 Hz [5,23], 600 Hz [25], 1000 Hz [20], and 2000 Hz [19].

Vicon [20,25], Optitrack [19] and Motus [5,23] systems, set to 120 Hz and 250 Hz, respectively, for Vicon, 200 Hz-Optitrack, 60 Hz-Motus were used to collect kinematic data in six works [5,19,20,23,25].

Lee et al. [24] used two accelerometers. One was placed on the subject, the other on the pendulum. A pendulum accelerometer (model 208CO3, PCB Piezotronics Inc, USA) was attached to the pendulum, and its signal determined the timing of pendulum release and impact. A participant-worn accelerometer (model 1356a16, PCB Piezotronics Inc. Inc, USA) was attached to the participant's dorsal surface at the L5S1 level, to detect temporal pelvic motion events.

Matjacić et al. [21] used multi-purpose rehabilitation frame (MRF). The MRF was used to generate perturbations for a standing subject. It consists of two 2-dof and two 1-dof rotational joints, two vertical support rods and a bracing system. Subjects stood with each foot on an aluminum block that constrained the position and orientation of their feet. Torque impulses, delivered by two hydraulic motors through a bracing system which was put around the subjects' pelvis, were used to induce perturbations.

Many authors [19,20,24,25] chose to measure lower-extremity muscle activity. All of them, with the exception of Zhu et al. [20], chose to measure activity on two lower limbs. The activity of the tibialis anterior (TA) muscle was recorded in each paper except Inacio et al. [25]. The authors additionally chose muscles such as: GMed, RF, ST, IL, GMax, AM [20]; GMed, TFL, ADD [25]; MG, RF, BF, GMed, EO, RA, ES [24]. In contrast, only Zhu et al. [20] included mechanomyogram data beside the EMG signal. The frequency of data collection was 2000 Hz [19,20], 1500 Hz [25], and 1000 Hz [24].

Ankle responses were included in publications by Rietdyk et al. [15], Zhu et al. [20], and Luchies et al. [19]. The knee joint was omitted by most authors and was only included in the study by Zhu et al. [20]. The pelvis and hip joints were mentioned in three reports [15,20,24]. CoP and CoM movements were the most frequently analyzed, as they were considered in four publications [5,20,24,25].

3.4. Description of the Sequence of Responses to Mediolateral Perturbations

The postural responses to mediolateral perturbations were obtained and successfully organized as a function of time. EMG activity appears earlier with serve perturbations. All studies show that the activity of the analyzed muscles appears earlier with severe perturbations than with weak ones.

Appendix A (Table A1) details the postural responses at the ankle-joint to medial variations in equilibrium. Their origin is associated with a change in the angular position of the calf in the transverse plane. It usually occurs 80 ms after the onset of the perturbation force. At this point, the dorsiflexors are activated, and after a slight delay, at 160.5 ms, the plantar flexors are involved. Depending on the severity of the perturbation, the activation of the previously mentioned muscles lasts up to about 250 ms. At lower perturbation intensities, activation of the relevant muscles may be delayed and begin around 400 ms. When strong perturbations occur, peak muscle activities are between 680 ms and 1300 ms. For moderate perturbations, this is between 530 ms and 1203 ms, and for small ones from 458 ms to 1274 ms. Thus, in most cases, the muscles reach maximum EMG activity between 530 ms and 1275 ms [20].

Responses occurring at the knee joint have only been considered in the study by Zhu et al. [20]. First, there are extension torques which occur between 100 ms and 200 ms, depending on the magnitude of the perturbation. Next, flexion torques are observed. The onset of the mechanomyography (MMG) signal of the knee joint flexors and extensors is recorded between 190 ms and 290 ms. Although with a small magnitude of the perturbation, MMG from the knee joint flexors appears around 380 ms. In the first case, the onset of activity in the knee extensors is observed at 250 ms. In the second, the activation of the knee flexors or extensors is observed at 600 ms. The peak of this muscle activity is up to 1250 ms. Muscle activity affects joint positioning and torques, which achieve their maximum values up to 1750 ms. A detailed description of the response of the knee joint to mediolateral perturbations appears in Appendix A (Table A2).

The behavior of the pelvis is closely linked with the hip joint. From 55 ms, hip torques and almost simultaneous (from 65 ms) angular position of the thigh changes are observed [15]. MMG activity of the hip extensors is visible from the very beginning, from approximately 50 ms, but the first EMG recordings appear at around 100 ms [20]. The first in-sequence EMG recordings come from the muscles responsible for hip abduction [20]. Maximum MMG activity is up to 1050 ms, and peak EMG activity is up to 1200 ms. It is worth noting that motion in the hip joint is not uniplanar. It consists of adduction and abduction, but also flexion and extension motions. Depending on the magnitude of the perturbation, the change in the hip flexion angle occurs in as little as 100 ms. In 450 ms, the hip extension angle changes [20]. Appendix A (Table A3) provides a detailed description of the responses of the pelvis and hips to lateral balance perturbations.

The joint and body segment responses described above are all reflected in CoP and CoM displacements. The first CoM motions, like those of the ankle joint, occur approximately 80 ms after perturbation initiation [20]. On the other hand, most reactions occur from 95 ms onward. The CoP displacement follows CoM motion. When the perturbation is small, CoP and CoM displacements occur later than during large ones [20]. It is also worth noting that when a perturbing force is applied to the shoulder girdle, CoP motion begins after 157 ms (medial–lateral displacement) and 191 ms (anterior–posterior displacement), respectively [24]. CoP and CoM motions as a function of time are shown in Appendix A (Table A4).

4. Discussion

The issue of postural responses to mediolateral perturbations in healthy, young people in the standing position is not often addressed in literature, as indicated by the inclusion of this review of only eight studies from 1999 to June 2022. More often mentioned are studies comparing young people with older adults [23,27], or postural stability during gait [28–30]. This review aimed to summarize and update information on how healthy, young adults deal with mediolateral perturbations in a standing position, including reviewing the methodology, outcomes, and sequence of responses in these studies. In this regard, analysis of the research methodology and results was conducted.

4.1. An Overview of the Research Methodology

In the current literature, there is no complete consensus on the methodology for conducting studies of this type and how to collect data. Only three authors [19,20,25] decided to conduct a complex analysis that included kinematic parameters from three-dimensional motion-capture systems, ground reaction forces, and EMG signals. However, the most commonly used duo is the measurement of ground reaction forces combined with EMG [19,20,24,25]. It makes it possible to observe how perturbation-induced CoP displacement affects muscle activity.

A literature review shows that pulling [19–21,23,25] is a more frequently selected method for inducing perturbations during free-standing, than pushing [15,24]. This inconsistency cannot be explained based on the authors' argumentation, since they did not specify why they chose this type of perturbation rather than another. However, it is

reasonable to assume that this was for practical reasons since it is easier to construct a pulling system than a pushing one. It remains unclear whether the responses to pulling and pushing are the same, because, to date, no work has appeared utilizing both mechanisms to induce perturbations, which leaves space for further research.

The situation is similar with the place of perturbation application. Most authors chose the pelvis [5,15,19–21,23,25], but none justified this decision. One may think it was motivated by its proximity to the body's center of gravity. However, the exact differences in responses to perturbations applied to the pelvis compared to the shoulder girdle are still unknown, and hence, it is not known whether they can be used interchangeably. Rietdyk et al. [15] were the only ones to conduct a study of the pelvic and shoulder rim perturbation responses. Perturbations were induced by pushing. For each condition, a different force (108.7 $\pm$ 1.22 N vs. 122 $\pm$ 1.3 N) was used for the shoulder and pelvis, respectively. They showed that displacements of body segments were by motion from proximal to distal. After shoulder and pelvis perturbations, the trunk had the highest excursion and moved before the lower extremities. The direction of trunk motion depended on its location. When the perturbation was at shoulder level, both the legs and the torso moved in the same direction as the perturbation, before returning to the vertical position. In response to the pelvic perturbation, the legs moved in the same direction as the disturbance, while the trunk moved on the opposite side. In addition, significantly higher CoP shifts were observed for perturbations at the pelvic level relative to the shoulders (12.5 $\pm$ 0.2 cm vs. 11.6 $\pm$ 0.2 cm). CoM displacements were much smaller in magnitude (6.4 $\pm$ 0.2 cm vs. 6 $\pm$ 0.1 cm) for the pelvis and shoulders, respectively. Based on this single paper, one may see that there are differences in reactions between pushing via pelvic and shoulder girdles. These findings may use confirmation by other authors. Yet, the question remains about the existence of such differences to pulling perturbations. The somatosensory system response may be different for pulls and pushes. In the reviewed papers, pulls occurred mainly with waist belts, which acted over a large area (the waist circumference). Pushes, on the other hand, were targeted and performed with a stick or bar. In our opinion, this stimulus setting would be significant in responding to this type of perturbation. However, none of the authors discussed this phenomenon. The neuronal activity of the left and right hemispheres of the brain is known to influence whether the left or right hand is dominant over the other [31]. Therefore, a similar relationship seems likely to occur when responding to induced perturbations. Unfortunately, none of the authors examined differences between responses to perturbations on the dominant and non-dominant (left/right) sides. In two papers [20,24], the authors mentioned the participants' dominant limb but only as an element of group characteristics. Since there are reports [31] in which the authors mention the faster response of the dominant side relative to the non-dominant side, it would be worthwhile to investigate whether responses to perturbations from the dominant lower limb differ from those of the non-dominant side. At this point, it is worth mentioning the paper of de Graaf et al. [32]. The authors showed that the activation of the sensorimotor network during the preparation of voluntary motor acts depends on whether motor perturbations are expected. When external forces are likely to disrupt movements in progress, the primary sensorimotor areas must be ready to respond to perturbations as quickly as possible to achieve the intended purpose of the motion under performance. Therefore, it is crucial to consider a number of factors when evaluating responses to perturbations. These include their predictability, location, direction, and strength. As this review shows, the authors considered only some of them, which influenced a variety of research methodologies.

Despite differences in study methodology, the comparison of the result unveiled interesting findings. Noteworthy is the use by Zhu et al. [20] of synchronous measurement of MMG and EMG signals, which is novel in this field of research. The authors showed that the detected MMG initial delays were earlier than those for the EMG signals, which did not follow the temporal sequence by which the initiation of electrical activity, measured by EMG, should precede the initiation of muscle twitching measured, by MMG [33,34]. It

indicates that the detected rapid response of MMG signals may not be generated by active and voluntary muscle contraction, but instead by passive and involuntary muscle motion, following the perturbation of the waist pull. In addition, this was supported by the earlier latency of the MMG initial signal generated by the hip muscles, which were closest to the place of perturbation. This very well corresponds with the findings of Rietdyk et al. [15]. They noticed that the joint torque onset occurred too early, suggesting that it is not based on feedback-initiated muscle activation, but rather due to muscle and tissue stiffness. Such stiffness-based balance control is straightforward, because it does not directly involve the CNS and sensory systems and provides, in theory, an immediate and appropriate corrective response.

In response to mediolateral perturbations, consideration of the ankle and hip joints, but not the knee joint, was adopted. This is likely due to the knee joint axis being parallel to the direction of the precipitating forces. In a 2019 publication, Lee et al. [24] described the up-and-down motion of the pelvis occurring during a shoulder thrust. It was not, however, explained why that displacement occurred. The only publication incorporating knee joint measurements was in 2022, where Zhu et al. [20] collected data on the angular position, torques, and EMG and MMG activity of the extensors and flexors of this joint. In doing so, they observed that this joint also contributes to recovery. However, it is worth noting that the levels of change recorded for the values of torques or angles were small and characterized by large standard deviations (Appendix A, Table A2). It may cause many authors to disregard this joint when analyzing the body's response to perturbations.

4.2. Timeline of Response Sequences to Mediolateral Perturbations

An important part of this study was an attempt to create a timeline on which individual postural responses to mediolateral perturbations were presented as a function of time. Due to the small number of available publications on the subject, divisions by weight, height, gender, or the site of the perturbation were abandoned, as the goal was to capture general regularities characterizing the response to this type of perturbation. Based on the Rietdyk et al. [15] paper, it can be concluded that the first reactions to ML perturbations at the pelvic level occur at the hip joints (55 ms). There is a change in angular alignment followed by torques. It is noteworthy that the MMG activity of the hip joint abductors occurs at the very beginning, but the EMG activity is noted with a longer delay. This phenomenon is repeated for other muscle groups in the later stages of the postural response [20]. The cause is most likely the secondary movement of the muscle bellies, due to angular changes in the nearby joint. That is in line with the findings of Rietdyk et al. [15]. They suggest that the initial response to perturbations is due to the flexibility of extra-articular structures rather than muscle activation.

The response of the ankle joints is observed after the initial displacement of the hip joints. As the onset of CoP displacement is observed (86.7 ms) and immediately afterward (93.1 ms), the CoM motion, the first muscles become activated, as evidenced by the EMG signal from the hip extensors, dorsiflexors, and plantar flexors. Subsequently, there is a noticeable movement in the knee joint, which, at approximately 200 ms, manifests itself in the presence of torques, as well as MMG activity, but no EMG signals [20]. Moreover, in this temporal region, Lee et al. [24] observed an up-down movement of the pelvis, which might correlate with the motion at the knee joints. Perhaps this indicates an attempt to lower the center of gravity to recover faster. However, this hypothesis requires confirmation in future studies. As mentioned above, a series of postural reactions leads to a peak displacement of CoP, followed by CoM. Zhu et al. [20] showed that with small perturbations, this relationship occurs earlier than with larger ones. CoP and CoM reach extreme positions within 625–1000 ms, but this does not end positional responses, which last up to 1750 ms, according to the same authors.

Although there were no studies analyzing the preventing of stepping among the young, some data were acquired from comparative studies between the young and the elderly. Based on studies [5,23] in which subjects took a step to regain balance, it can be

concluded that the loaded side-step (LSS) strategy, which is preferably utilized by the young, allows faster initiation of step response than an unloaded crossover step (UCS). This appears to be especially relevant in the context of deciding if the step is needed or not. The decision to take a step is made by the CNS up to a maximum of 150 ms, allowing approximately 300–450 ms to initiate the step [4]. It is also worth noting, that this decision must be made by CNS before any EMG activity, i.e., before any active trial to restore equilibrium. In that case, the question arises on what basis CNS decides to take a step or not. In future studies, it might be worth investigating if there are differences in primal reactions to perturbations provoking stepping responses, and those that do not provoke one.

4.3. Study Limitation

This study has limitations, of which the main was a relatively small number of publications focused on the response to perturbations of young adults. It is worth noting that in the present study, we planned to include studies comparing the responses to mediolateral perturbations of healthy young adults and those with the disorders, since we could extract more information. However, based on the large methodological differences observed in studies involving young adults, we decided not to include another study group. Moreover, a small number of included publications resulted in the necessity of combining responses to mediolateral perturbations without a breakdown by gender, height, or weight. Additionally, due to differences in the methodology of included studies, their comparison was difficult and allowed for very general conclusions. As we mentioned earlier, it remains unclear whether the responses to pulling and pushing are the same. Although the perturbations associated with mechanical pushing and pulling appear to be the same, there may be other differences between them. In included studies, pulling was conducted via a harness. As a result, a larger contact surface with the subjects' bodies, and hence improved sensation from the skin surface, occurred. In contrast, pushing perturbations were provided to the smaller contact surface. Since tactile sensations are transmitted through myelinated Aβ fibers, their conduction velocity is high (approximately 80 m/s) [35]. In other words, the information gathered from surface receptors could potentially be used to modify the primary perturbation response. However, to date, no paper investigating this issue has appeared, which leaves space for further research.

5. Conclusions

Based on the review, it was observed that the issue of the body's response to mediolateral perturbations during free standing is open to further research. The reviewing process uncovered methodological inconsistencies in reviewed studies. The first one was the determination of the onset of perturbation. Some authors [21,24] defined it as the point at which the perturbation device began to move, but did not touch the subject. The main issue was how perturbations are triggered. The papers described responses to pushes and pulls at both pelvic and shoulder levels. However, none of the authors justified their choice. It is still unclear whether the postural responses to pulling and pushing are the same, even though pulling is more common. Moreover, authors more often chose perturbations at the pelvic level, yet differences between responses to pelvic and shoulder perturbations are unknown, to date. The issue of different magnitudes of perturbation forces was addressed in only one paper [20]; thus, the topic of reactions to different perturbation magnitudes remains open for investigation. Moreover, there were differences in data processing (filtering), that may have influenced the comparison of publications.

Regarding the timing of events in the perturbation response, the primary reaction appears to come from the hip joints. However, the knee joints also play a role in the postural response and may be involved in lowering CoM displacement, however, that issue needs further investigation. In mediolateral perturbation-type feedback responses, the fact that CoP motion precedes CoM translation may be important.

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Appendix A

Table A1. A detailed description of the temporal reactions to mediolateral perturbations in the ankle joint (AJ). The provided data are mainly from the paper [20], where: DF—dorsiflexors; PF—plantar flexors; ipsi.—ipsilateral; max pert.—maximum magnitude of the lateral perturbation.

Time [ms]	**80**	**87.1**	**85**	**90**	**95**	**97**	**110**
Event	Ipsi. angle change onset at pelvis push [15] Angle change onset in ankle joint opposite the perturbed side at pelvis push [15]	DF torque onset at max pert.	Torque onset in ankle joint opposite the perturbed side at pelvis push [15]	Angle change onset in AJ opposite the perturbed side at shoulder push [15] Ipsi. torque onset at pelvis push [15]	Ipsi. angle change onset at shoulder push [15]	Tibialis anterior EMG onset [19]	Ipsi. torque onset at shoulder push [15] Torque onset in ankle joint opposite the perturbed side at shoulder push [15]
Time [ms]	**135.5**	**160.5**	**163.6**	**172.3**	**175.6**	**190.1**	**191.2**
Event	DF torque onset at medial lateral pert.	PF angle change onset at minimal medial pert.	MMG onset and PF at maximal lateral pert.	MMG onset and PF at medium lateral pert.	MMG onset and DF at maximal lateral pert.	MMG onset and PF at medium lateral pert.	MMG onset and PF at maximal medial pert.
Time [ms]	**192.0**	**191.8**	**192.2**	**197.6**	**198.6**	**206.4**	**215.3**
Event	MMG onset and DF at maximal medial pert.	EMG onset and DF at maximal lateral pert.	EMG onset and DF at medium lateral pert.	MMG onset and DF at medium lateral pert.	Power absorbing onset at maximal lateral pert.	EMG onset and PF maximal lateral pert.	MMG onset and DF at medium medial pert.
Time [ms]	**230.6**	**232.5**	**241.8**	**250.1**	**250.9**	**272.9**	**279.6**
Event	MMG onset and PF at minimal lateral pert.	DF angle change onset at maximal medial pert.	EMG onset and DF at maximal medial pert.	DF torque onset at minimal lateral pert.	EMG onset and PF at medial lateral pert.	EMG onset and DF at medium medial pert.	EMG onset and DF at minimal lateral pert.
Time [ms]	**284.3**	**298.2**	**306.9**	**312.2**	**323.0**	**333.7**	**351.2**
Event	MMG onset and DF at minimal lateral pert.	MMG onset and PF at minimal medial pert.	DF torque onset at maximal medial pert.	MMG onset and DF at minimal medial pert.	Stepping foot lift of the plate [19]	Power generating onset at maximal medial pert.	EMG onset and PF at minimal lateral pert.
Time [ms]	**355.3**	**370.8**	**371.0**	**374.2**	**375.0**	**378.9**	**400.0**
Event	DF torque at medium medial pert.	PF torque onset at medium medial pert.	Power absorbing onset at maximal medial pert.	DF angle change onset at maximal lateral pert.	LSS step onset [4,5]	Power absorbing onset at minimal medial pert.	UCS step onset [4,5]
Time [ms]	**402.6**	**406.9**	**409.0**	**426.3**	**444.9**	**449.3**	**451.3**
Event	DF angle change onset at medium medial pert.	PF torque at maximal medial pert.	PF torque onset at minimal medial pert.	Power absorbing onset at minimal lateral pert.	DF angle change onset at minimal medial pert.	Power generating onset at medium medial pert.	EMG onset and DF at minimal medial pert.

Table A1. *Cont.*

Time [ms]	**455.3**	**458.4**	**461.0**	**502.1**	**504.6**	**505.0**	**519.1**
Event	Power generating onset at minimal lateral pert.	Peak of hip ext MG of DF at minimal lateral pert.	Stepping foot landing. Step length: 23.7 ± 11.9 cm; Heigth: 5.1 ± 2.0 cm; Step duration: 221 ± 175 ms [19].	Power generating onset at medium medial pert.	DF angle change onset at medium medial pert.	Power absorbing onset at medium lateral pert.	DF torque onset at minimal medial pert.
Time [ms]	**525.0**	**529.5**	**531.9**	**532.0**	**542.7**	**555.1**	**565.8**
Event	PF angle change onset at maximal medial pert.	Peak MMG of PF at minimal lateral pert.	Peak of hip ext. MG of DF at medial lateral pert.	PF angle change onset at medium medial pert.	DF angle change onset at minimal lateral pert.	Power generating onset at maximal lateral pert.	PF torque onset at minimal lateral pert.
Time [ms]	**574.8**	**588.4**	**601.8**	**605.9**	**607**	**608.3**	**612.9**
Event	Peak of hip ext. MG of PF at medial lateral pert.	Peak MMG of DF at minimal laterel pert.	Peak of hip ext. MG of PF at minimal lateral pert.	Peak MMG of PF at medial lateral pert.	Shank peak angle (−5.8 ± 0.2 deg) at pelvis push. [15]	Peak of hip ext. MG of DF at medium medial pert.	Power generating onset at minimal medial pert.
Time [ms]	**627.0**	**641.3**	**662.4**	**667.9**	**676.8**	**679.0**	**679.9**
Event	Peak of hip ext. MG of DF at minimal medial pert.	PF angle change onset at minimal lateral pert.	Power absorbing onset at medium medial pert.	Peak power absorbing (0.16 ± 0.11 W/kg) at maximal lateral pert.	Peak PF angle (0.6 ± 0.4 deg) at minimal medial pert.	Peak MMG of DF at medial lateral pert.	Peak of hip ext. MG of DF at maximal lateral pert.
Time [ms]	**696.7**	**698.2**	**700.0**	**711.0**	**742.3**	**749.4**	**758.4**
Event	PF angle change onset at maximal lateral pert.	Peak MMG of DF at minimal medial pert.	LSS step termination. Step duration 325 ms; lateral displacement and step clarence normalised by height: 0.16 cm [4,5].	Peak of hip ext. MG of PF at maximal lateral pert.	Peak MMG of PF at minimal medial pert.	Peak power absorbing (0.01 ± 0.01 W/kg) at minimal medial pert.	Peak MMG of DF at medium medial pert.
Time [ms]	**771.7**	**778.4**	**780.6**	**784.4**	**787.2**	**806.4**	**808.2**
Event	EMG onset of PF at medium medial pert.	Peak PF torque (0.12 ± 0.05 Nm/kg) at minimal medial pert.	Peak power generating (0.03 ± 0.02 W/kg) at medium lateral pert.	Peak power absorbing (0.02 ± 0.02 W/kg) at medium lateral pert.	Peak power absorbing (0.01 ± 0.01 W/kg) at minimal lateral pert.	PF angle change onset at medium lateral pert.	EMG onset at PF at maximal medial pert.
Time [ms]	**816.2**	**827.9**	**836.4**	**849.3**	**868.2**	**870.4**	**881**
Event	Peak PF torque (0.16 ± 0.06 Nm/kg) at medium medial pert.	Peak DF torque (0.41 ± 0.2 Nm/kg) at maximal lateral pert.	Peak DF torque at medium lateral pert. 0.22 ± 0.09 Nm/kg	EMG onset at PF at minimal medial pert. Peak MMG of PF at medium medial pert.	Peak DF angle (3.1 ± 1.6 deg) at medium medial pert.	PF torque onset at medium lateral pert.	Shank peak angle (−2.6 ± 0.2 deg) at shoulder push [15]
Time [ms]	**902.8**	**912.9**	**915.5**	**929.2**	**931.6**	**948.4**	**952.3**
Event	Peak power generating (0.01 ± 0.01 W/kg) at minimal lateral pert.	Peak power generating (0.02 ± 0.01 W/kg) at medium medial pert.	Peak DF angle (1.6 ± 1.0 deg) at minimal medial pert.	Peak of hip ext. MG of DF at maximal medial pert.	Peak power absorbing (0.02 ± 0.01 W/kg) at medium medial pert.	Peak DF angle (8.4 ± 5.5 deg) at maximal lateral pert.	Peak PF torque (0.06 ± 0.04 Nm/kg) at minimal lateral pert.
Time [ms]	**952.7**	**963.5**	**981.2**	**998.4**	**1002.3**	**1014.6**	**1025.0**
Event	Peak DF angle (4.2 ± 2.3 deg) at maximal medial pert.	Peak power generating (0.01 ± 0.01 W/kg) at minimal medial pert.	Peak MMG of PF at maximal lateral pert.	Peak PF angle (1.7 ± 1.0 deg) at medium medial pert.	Peak MMG of DF at maximal lateral pert.	Peak DF torque (0.13 ± 0.08 Nm/kg) at minimal lateral pert.	UCS step termination. Step duration 625 ms; lateral displacement and step clearance normalised by body heigth: 0.34 cm [4,5]

Table A1. *Cont.*

Time [ms]	**1033.1**	**1067.8**	**1076.7**	**1090.0**	**1116.4**	**1130.5**	**1139.8**
Event	PF torque onset at maximal lateral pert.	Peak MMG of PF at maximal medial pert.	Peak PF angle (98.2 ± 4.6 deg) at maximal medial pert.	Peak power generating (90.04 ± 0.03 W/kg) at maximal medial pert.	Peak PF torque 90.18 ± 0.09 Nm/kg) at maximal medial pert.	Peak MMG of DF at maximal medial pert.	Peak power generating (0.10 ± 0.08 W/kg) at maximal lateral pert.
Time [ms]	**1152.3**	**1174.4**	**1190.9**	**1203.9**	**1211.5**	**1244.9**	**1274.2**
Event	Peak PF angle (0.9 ± 0.3 deg) at minimal lateral pert.	Peak DF angle (2.9 ± 2.3 deg) at medium lateral pert.	Peak PF torque (0.06 ± 0.05 Nm/kg) at medium lateral pert.	Peak of hip ext. MG of PF at medium medial pert.	Peak DF angle (1.4 ± 0.7 deg) at minimal lateral pert.	Peak DF torque (0.13 ± 0.07 Nm/kg) at medium medial pert.	Peak of hip ext. MG of PF at minimal medial pert.
Time [ms]	**1294.0**	**1299.9**	**1302.1**	**1310.0**	**1318.4**	**1337.7**	**1396.1**
Event	Peak PF angle (1.1 ± 0.5 deg) at medium lateral pert.	Peak of hip ext. MG of PF at maximal medial pert.	Peak PF torque (0.14 ± 0.08 Nm/kg) at maximal lateral pert.	Peak DF torque (0.17 ± 0.10 Nm/kg) at maximal medial pert.	Peak DF torque (0.11 ± 0.09 Nm/kg) at minimal medial pert.	Peak power absorbing (0.06 ± 0.03 W/kg) at maximal medial pert.	Peak PF angle (1.3 ± 0.8 deg) at maximal lateral pert.

Table A2. A detailed description of the temporal reactions to mediolateral perturbations in the knee joint based on [20], where: ext.—knee extensors; flex.—knee flexors.

Time [ms]	**64.9**	**161.0**	**187.0**	**192.1**	**200.7**	**203.4**	**205.2**
Event	Knee ext. torque onset at medium medial pert.	Knee ext. torque onset at maximal medial pert.	Knee ext. MMG onset at maximal lateral pert.	Knee flex. MMG onset at maximal medial pert.	Power absorbing onset in knee at maximal lateral pert.	Knee flex. MMG onset at medial lateral pert.	Knee flex. MMG onset at maximal lateral pert.
Time [ms]	**213.1**	**215.7**	**224.5**	**225.0**	**225.2**	**226.7**	**244.9**
Event	Knee ext. torque onset at minimal medial pert.	Knee flex. torque onset at medial lateral pert.	Knee ext. MMG onset at medial lateral pert.	Knee flex. torque onset at maximal lateral pert.	Knee flex. MMG onset at medium medial pert.	MMG onset knee ext. at maximal medial pert.	Knee flex. torque onset at minimal lateral pert.
Time [ms]	**245.1**	**238.7**	**289.6**	**303.8**	**308.5**	**326.7**	**331.7**
Event	Knee ext. MMG onset at medium medial pert.	Angle change onset of knee flex. at maximal medial pert.	Knee flex. MMG onset at minimal lateral pert.	Knee ext. MMG onset at minimal lateral pert.	EMG onset of knee ext. at maximal lateral pert.	Knee ext. MMG onset at minimal medial pert.	Angle change onset of knee flex. at medium medial pert.
Time [ms]	**334.2**	**334.3**	**336.9**	**351.7**	**354.9**	**358.8**	**363.8**
Event	EMG onset of knee flex. at maximal medial pert.	Angle change onset of knee flex. MID L pert.	Power absorbing onset in knee at medial lateral pert.	Power absorbing onset at medium medial pert.	Knee flex. MMG onset at minimal medial pert.	Angle change onset of knee flex. at minimal medial pert.	Power absorbing onset at maximal medial pert.
Time [ms]	**366.1**	**378.9**	**388.3**	**395.7**	**405.9**	**411.0**	**429.8**
Event	Angle change onset of knee ext. at maximal medial pert.	Power absorbing onset in knee at minimal medial pert.	Power generating onset at maximal lateral pert.	Power absorbing onset at minimal lateral pert.	Power generating onset at maximal medial pert.	EMG onset of knee ext. at medial lateral pert.	EMG onset of knee flex. at medium medial pert.
Time [ms]	**464.3**	**470.1**	**482.6**	**487.3**	**494.8**	**495.0**	**505.3**
Event	Angle change onset of knee flex. at maximal laterel pert.	Angle change onset of knee flex. at mimimal lateral pert.	Angle change onset of knee ext. at maximal lateral pert.	EMG onset of knee ext. at maximal medial pert.	Knee ext. torque onset at medial lateral pert.	Angle change onset of knee ext. at minimal medial pert.	Power generating onset at medial lateral pert.
Time [ms]	**507.1**	**532.4**	**553.6**	**562.9**	**566.2**	**574.3**	**577.9**
Event	Angle change onset of knee ext. at minimal lateral pert.	EMG onset of knee flex. at maximal lateral pert.	Knee ext. torque onset at minimal lateralpert.	Knee ext. torque onset at maximal lateral pert.	Angle change onset of knee ext. at medium lateral pert.	Power generating onset at minimal lateralpert.	Peak of knee flex. torque (0.27 ± 0.15 Nm/kg) at medial lateral pert.

Table A2. *Cont.*

Time [ms]	**582.0**	**586.3**	**596.1**	**623.1**	**651.3**	**652.2**	**600.8**
Event	Peak of hip flex. torque (0.18 ± 0.10 Nm/kg) at minimal lateral pert.	Peak power absorbing (0.05 ± 0.03 W/kg) at medial lateral pert.	Knee ext. MMG onset at minimal lateral pert.	Peak MMG of knee flex. aminimal medial pert.	Peak MMG of knee flex. at minimal lateral pert.	Peak MMG of knee flex. at medial lateral pert.	EMG onset knee ext. at minimal lateral pert.
Time [ms]	**627.1**	**653.2**	**659.0**	**679.0**	**693.4**	**697.8**	**699.5**
Event	EMG onset knee flex. at minimal medial pert. Peak power absorbing (0.02 ± 0.01 W/kg) at minimal lateral pert.	Peak MMG of knee ext. at minimal medial pert.	Peak of EMG of knee ext. at medial lateral pert.	Power generating onset at medium medial pert.	Peak power absorbing (0.17 ± 0.20 W/kg) at maximal lateral pert.	Peak MMG of knee ext. at medial lateral pert.	Peak of knee flex. angle (1.2 ± 1.1 deg) at medial lateral pert.
Time [ms]	**724.7**	**730.4**	**736.7**	**738.9**	**739.9**	**756.4**	**762.9**
Event	Power generating onset at minimal medial pert.	EMG onset knee flex. at medial lateral pert.	EMG onset of knee ext. at medium medial pert.	Peak EMG of knee ext. at maximal lateral pert.	Peak EMG of knee ext. at medium medial pert.	Peak EMG of knee ext. at minimal lateral pert.	Peak MMG of knee flex. at medium medial pert.
Time [ms]	**769.6**	**770.4**	**772.0**	**783.3**	**808.9**	**811.9**	**812.6**
Event	Peak power absorbing (0.01 ± 0.01 W/kg) at minimal medial pert.	Peak of hip ext torque (0.14 ± 0.07 Nm/kg) at minimal medial pert.	Peak MMG of knee ext. at medium medial pert.	Peak of knee flex. angle (5.6 ± 3.0 deg) at medium medial pert.	Peak power absorbing (0.03 ± 0.01 W/kg) at medium medial pert.	Peak of hip ext. torque (0.21 ± 0.10 Nm/kg) at medium medial pert.	Peak power generating (0.02 ± 0.01 W/kg) at minimal lateral pert.
Time [ms]	**818.5**	**844.8**	**856.7**	**862.7**	**876.4**	**890.2**	**891.7**
Event	EMG onset of knee flex. at minimal lateral pert.	EMG onset of knee ext. at minimal medial pert.	Peak EMG of knee flex. at medium medial pert.	Peak of knee flex. angle (2.4 ± 1.6 deg) at minimal medial pert.	Peak power generating (0.04 ± 0.02 W/kg) at medial lateral pert.	Peak of knee flex. angle (0.7 ± 0.3 deg) at minimal lateral pert.	Peak MMG of knee ext. at maximal lateral pert.
Time [ms]	**902.3**	**912.9**	**919.1**	**950.8**	**953.8**	**960.2**	**969.1**
Event	Peak of hip flex. angle (15.1 ± 8.6 deg) at maximal medial pert.	Angle ext. change onset of at medium medial pert.	Peak power generating (0.20 ± 0.23 W/kg) at maximal lateral pert.	Peak EMG of knee flex. at medial lateral pert.	Peak of knee flex. torque (0.48 ± 0.24 Nm/kg) at maximal lateral pert.	Knee flex. torque onset at minimal medial pert.	Peak EMG of knee flex. at minimal medial pert.
Time [ms]	**972.8**	**976.7**	**982.9**	**992.9**	**1020.5**	**1011.4**	**1027.6**
Event	Peak of knee flex. angle (5.4 ± 4.6 deg) at maximal lateral pert.	Peak power generating (0.01 ± 0.01 W/kg) at minimal medial pert.	Peak MMG of knee flex. at maximal medial pert.	Peak power generating (0.09 ± 0.07 W/kg) at maximal medial pert.	Peak EMG of knee ext. at maximal medial pert.	Peak knee ext. angle (0.8 ± 0.9 deg) at minimal medial pert.	Peak MMG of knee flex. at maximal lateral pert.
Time [ms]	**1046.0**	**1065.2**	**1082.1**	**1069.6**	**1078.5**	**1109.4**	**1122.1**
Event	Peak knee ext. angle (2.4 ± 2.0 deg) at maximal lateral pert.	Peak MMG of knee ext. at maximal medial pert.	Peak of hip ext. torque (0.24 ± 0.11 Nm/kg) at maximal medial pert.	Peak power generating (0.02 ± 0.01 W/kg) at medium medial pert.	Peak EMG of knee flex. at minimal lateral pert.	Peak EMG of knee flex. at maximal medial pert.	Peak EMG of knee ext. at minimal medial pert.
Time [ms]	**1184.0**	**1192.4**	**1193.7**	**1230.3**	**1235.9**	**1266.2**	**1262.9**
Event	Peak of knee ext. angle (1.9 ± 1.5 deg) at maximal medial pert.	Peak of knee ext. angle (1.4 ± 1.4 deg) at minimal lateral pert.	Peak EMG of knee flex. at maximal lateral pert.	Peak of knee ext. angle (1.8 ± 0.8 deg) at medial lateral pert.	Flex. torque onset at medium medial pert.	Peak of hip ext. torque (0.09 ± 0.06 Nm/kg) at minimal lateral pert.	Peak power absorbing (0.07 ± 0.06 W/kg) at maximal medial pert.
Time [ms]	**1344.1**	**1348.7**	**1353.2**	**1463.8**	**1470.8**	**1538.5**	**1762.4**
Event	Peak knee flex. torque (0.08 ± 0.06 Nm/kg) at minimal medial pert.	Peak of hip ext. torque (0.13 ± 0.09 Nm/kg) at maximal lateral pert.	Peak of knee ext. angle (1.4 ± 1.0 deg) at medium medial pert.	Peak of knee ext. torque (0.13 ± 0.07 Nm/kg) at medial lateral pert.	Knee flex. torque onset at maximal medial pert.	Peak of knee flex. torque (0.10 ± 0.09 Nm/kg) at medium medial pert.	Peak of knee flex. torque (0.22 ± 0.14 Nm/kg) at maximal medial pert.

Table A3. A detailed description of the temporal reactions to mediolateral perturbations in the pelvis and hip joint (HJ). The provided data are mainly from the paper [20], where: ABD—hip abductors, ADD—hip adductors.

Time [ms]	**55**	**65**	**62.3**	**63.3**	**65.4**	**70**	**75**
Event	Ipsi. torque onset at pelvis push [15]	Ipsi. angle change onset at pelvis push [15]	MMG onset of hip ABD at maximal lateral pert.	Hip ABD torque onset at medium medial pert.	MMG onset of hip ABD at maximal medial pert.	Torque onset change in HJ opposite the perturbed side at shoulder and pelvis push [15]	Angle change onset in HJ opposite the perturbed side at shoulder and pelvis push [15]
Time [ms]	**78.2**	**80**	**83.4**	**86.7**	**86.9**	**87.3**	**89.4**
Event	MMG onset of hip ABD at medium medial pert.	Ipsi. angle change onset at shoulder push [15]	Hip ABD torque onset at maximal medial pert.	MMG of hip flex. onset at maximal medial pert.	Hip flex. torque onset at medium medial pert.	MMG of hip ABD onset at minimal lateral pert.	MMG of hip ABD onset at medial lateral pert.
Time [ms]	**93.6**	**103.7**	**110**	**115.6**	**117.4**	**123.4**	**124.4**
Event	MMG of hip flex. onset at maximal lateral pert.	MMG of hip ext. onset at medium medial pert.	Ipsi. torque onset at shoulder push [15]	MMG of hip flex. onset at medial lateral pert.	Hip ABD torque onset at minimal medial pert.	MMG of hip flex. onset at medium medial pert.	Angle of hip flex. change onset at maximal lateral pert.
Time [ms]	**130.6**	**131.3**	**135.7**	**143.9**	**148.9**	**151.2**	**154.2**
Event	MMG of hip ext. onset at maximal lateral pert.	Power generation torque at medium medial pert.	Power generation torque at minimal medial pert.	MMG onset of hip ABD at minimal medial pert.	MMG of hip ext. onset at minimal lateral pert.	Hip ADD torque onset at maximal lateral pert.	MMG of hip ext. onset at medial lateral pert.
Time [ms]	**159.1**	**161.2**	**163.4**	**169.1**	**173.1**	**179.4**	**176.9**
Event	MMG of hip ADD onset at medium medial pert.	Hip ADD torque onset at medial lateral pert.	EMG onset of hip ABD at maximal lateral pert.	MMG onset of hip ADD at maximal medial pert.	MMG of hip flex. onset at minimal lateral pert.	MMG of hip ADD onset at maximal lateral pert.	MMG of hip ext. onset at minimal medial pert.
Time [ms]	**177.9**	**180.6**	**181.2**	**183.6**	**183.7**	**193.0**	**193.4**
Event	Power generation torque at maximal medial pert.	EMG of hip ABD onset at medial lateral pert.	Hip flex. torque onset at maximal medial pert.	MMG of hip ext. onset at maximal medial pert.	Hip ext. torque onset at medial lateral pert.	Hip flex. angle change onset at medial lateral pert.	Hip flex. angle change onset at medium medial pert.
Time [ms]	**195.6**	**197.9**	**198.9**	**206.4**	**220.1**	**224.2**	**225.6**
Event	Hip ADD torque onset at minimal lateral pert.	MMG of hip ADD onset at medial lateral pert.	Hip flex. Torque onset at minimal medial pert.	Hip ext. torque onset at minimal lateral pert.	Hip ABD angle change onset at minimal medial pert.	EMG of hip ABD onset at minimal lateral pert.	EMG of hip flex. onset at maximal lateral pert.
Time [ms]	**226.6**	**234.2**	**233.17**	**233.57**	**235.61**	**246.2**	**256.7**
Event	MMG of hip flex. onset at minimal medial pert.	MMG of hip ADD onset at minimal lateral pert.	Pelvic up-down motion [24]	Pelvic ML motion [24]	Pelvic AP motion [24]	MMG of hip ADD onset at minimal medial pert.	EMG of hip ABD onset at maximal medial pert.
Time [ms]	**258.0**	**263.7**	**263.8**	**270.3**	**272.2**	**272.4**	**284.1**
Event	Power absorbing onset at minimal lateral pert.	Hip ADD angle change onset at minimal lateral pert.	Hip ABD angle change onset at medium medial pert.	Hip flex. angle chang onset at maximal medial pert.	EMG of hip flex. Onset at medial lateral pert.	Hip ABD angle change onset at maximal medial pert.	EMG of hip ext. onset at maximal lateral pert.
Time [ms]	**301.4**	**301.6**	**306.4**	**309.0**	**314.9**	**315.1**	**326.4**
Event	Hip flex. angle change onset at minimal medial pert.	EMG of hip ABD onset at medium medial pert.	Hip ext. torque onset at maximal lateral pert.	Power generation onset at maximal lateral pert.	EMG of hip ext. onset at medial lateral pert.	Power generation onset at medial lateral pert.	EMG of hip ADD onset at maximal medial pert.
Time [ms]	**332.6**	**342.5**	**347.7**	**363.0**	**370.9**	**388.9**	**389.3**
Event	EMG of hip ADD onset at minimal medial pert.	Power absorbing onset at medial lateral pert.	Hip ADD angle change onset at medial lateral pert.	Power absorbing onset at maximal lateral pert.	Hip ADD angle change onset at maximal lateral pert.	Peak power generating at minimal medial pert.	EMG of hip flex. onset at maximal medial pert.

Table A3. *Cont.*

Time [ms]	**415.4**	**418.6**	**430.8**	**444.0**	**473.4**	**479.6**	**480.4**
Event	Peak of hip ABD torque (0.10 ± 0.07 Nm/kg) at medium medial pert.	Hip flex. torque onset at medial lateral pert.	Hip flex. angle change onset at minimal lateral pert.	EMG of hip ADD onset at medium medial pert.	EMG of hip ext. onset at minimal lateral pert.	EMG of hip flex. onset at minimal lateral pert.	Hip ext. angle change onset at maximal medial pert.
Time [ms]	**480.6**	**490.9**	**504.7**	**510.4**	**513.9**	**523.5**	**523.7**
Event	Peak MMG of hip flex. at minimal lateral pert.	Peak MMG of hip ext. at minimal medial pert.	Peak power generating (0.03 ± 0.02 W/kg) at medium medial pert.	Peak of EMG of hip ABD at medial lateral pert.	Peak MMG of hip ABD at minimal lateral pert.	Peak of EMG of hip ABD at minimal lateral pert.	Peak of hip ABD torque (0.09 ± 0.06 Nm/kg) at minimal medial pert.
Time [ms]	**536.1**	**542.0**	**544.4**	**544.9**	**547.1**	**573.8**	**555.0**
Event	Peak MMG of hip flex. at medial lateral pert.	Peak power absorbing (0.05 ± 0.02 W/kg) at minimal lateral pert.	Peak of EMG of hip flex. at medial lateral pert.	Peak of EMG of hip ext. at medial lateral pert.	EMG of hip ADD onset at minimal lateral pert.	Peak power absorbing (0.15 ± 0.08 W/kg) at medial lateral pert.	EMG of hip ABD onset at minimal medial pert.
Time [ms]	**562.1**	**581.9**	**586.6**	**589.1**	**597.6**	**645.4**	**599.7**
Event	Peak of EMG of hip ADD at minimal medial pert.	EMG of hip ext. onset at medium medial pert.	Power absorbing onset at maximal medial pert.	EMG of hip ADD onset at maximal lateral pert.	EMG onset of hip flex. at medium medial pert.	Peak angle of hip flex. (2.1 ± 1.2 deg) at medial lateral pert.	Peak of EMG of hip ext. at minimal lateral pert.
Time [ms]	**601.9**	**604.7**	**607.6**	**611.1**	**615.6**	**621.3**	**624.3**
Event	Angle change onset of hip ABD at maximal lateral pert.	Peak of hip ext torque (0.40 ± 0.23 Nm/kg) at medial lateral pert.	Peak MMG of hip flex. at minimal medial pert.	EMG of hip ADD onset at medial lateral pert.	Peak of hip ADD torque (0.37 ± 0.11 Nm/kg) at minimal lateral pert.	Peak angle of hip ABD (2.4 ± 1.7 deg) at minimal medial pert.	Peak of hip ADD torque (0.79 ± 0.19 Nm/kg) at medial lateral pert.
Time [ms]	**626.9**	**634.4**	**637.0**	**644.4**	**651.1**	**656.4**	**663.7**
Event	Hip ADD torque onset at maximal medial pert.	Hip flex. torque onset at maximal lateral pert.	Peak of EMG of hip flex. at minimal lateral pert.	Peak MMG of hip ADD at minimal medial pert.	Peak MMG of hip ABD at minimal medial pert.	Peak of hip ABD torque (0.15 ± 0.09 Nm/kg) at maximal medial pert.	Peak of EMG of hip flex. at maximal lateral pert.
Time [ms]	**664.3**	**664.4**	**664.9**	**671.1**	**678.4**	**678.6**	**681.9**
Event	Peak of EMG of hip ABD at maximal lateral pert.	Peak of hip ADD torque (1.01 ± 0.19 Nm/kg) at maximal lateral pert.	Peak ABD hip angle (3.2 ± 2.1 deg) of at medium medial pert.	Peak hip flex. torque (0.19 ± 0.10 Nm/kg) at minimal medial pert.	Peak MMG of hip ABD at medial lateral pert.	Flex. hip torque onset at minimal lateral pert.	Hip ABD angle change onset of at medial lateral pert.
Time [ms]	**684.6**	**690.1**	**696.8**	**697.7**	**698.6**	**703.6**	**707.7**
Event	Peak MMG of hip ADD at minimal lateral pert.	Peak MMG of hip ext. at medial lateral pert.	Peak MMG of hip ABD at medium medial pert.	Peak of EMG of hip ADD at minimal lateral pert. 4	Peak MMG of hip flex. at medium medial pert.	Peak power generating (0.08 ± 0.04 W/kg) at medial lateral pert.	Peak MMG of hip ext. at medium medial pert.
Time [ms]	**716.1**	**716.3**	**719.0**	**726.5**	**728.8**	**730.8**	**732.4**
Event	Peak of hip ext. torque (0.24 ± 0.15 Nm/kg) at minimal lateral pert.	Peak of EMG of hip ext. at maximal lateral pert.	Peak MMG of hip ABD at maximal lateral pert.	Peak of EMG of hip ADD at medium medial pert.	EMG onset of hip ext. at maximal medial pert.	Peak angle of hip ADD (2.2 ± 1.2 deg) at minimal lateral pert.	Power generating onset at minimal lateral pert.
Time [ms]	**739.9**	**749.7**	**753.2**	**761.9**	**762.9**	**765.9**	**771.0**
Event	Peak of EMG of hip ext. at medium medial pert.	Peak angle of hip flex. (2.4 ± 1.2 deg) at minimal medial pert.	Peak power generating (0.21 ± 0.12 W/kg) at maximal lateral pert.	Peak power absorbing (0.26 ± 0.11 W/kg) at maximal lateral pert.	Peak power generating (0.13 ± 0.19 W/kg) at maximal medial pert.	Peak of hip flex. torque (0.27 ± 0.15 Nm/kg) at medium medial pert.	Peak angle of hip flex. (4.3 ± 2.3 deg) at medium medial pert.

Table A3. *Cont.*

Time [ms]	**772.9**	**773.0**	**787.1**	**793.1**	**807.2**	**807.9**	**810.1**
Event	Peak MMG of hip ADD at medium medial pert.	Peak of EMG of hip flex at medium medial pert.	Power absorbing onset at minimal medial pert.	Peak angle of hip ADD (2.9 ± 1.8 deg) at medial lateral pert.	Angle change onset of hip ext. at minimal lateral pert.	Power absorbing onset at medium medial pert.	Peak MMG of hip ADD at medial lateral pert.
Time [ms]	**818.7**	**834.2**	**835.9**	**842.7**	**842.9**	**844.7**	**856.4**
Event	Peak MMG of hip ADD at medial lateral pert.	EMG onset of hip ext. atminimal medial pert.	Angle change onset of hip ADD at maximal medial pert.	Peak MMG of hip flex. at maximal lateral pert.	Angle change onset of hip ADD at medium medial pert.	Peak MMG of hip ext. at minimal lateral pert.	EMG onset hip flex at minimal medial pert.
Time [ms]	**863.0**	**868.3**	**872.3**	**873.7**	**873.9**	**874.9**	**881.2**
Event	Peak MMG of hip ABD at maximal medial pert.	Hip ADD torque onset at medium medial pert.	Peak angle of hip flex. at minimal lateral pert. 0.9 ± 0.3 deg	Peak MMG of hip ADD at maximal medial pert.	Peak MMG of hip flex. at maximal medial pert.	Peak of hip ext torque (0.48 ± 0.31 Nm/kg) at maximal lateral pert.	Angle change onset of hip ext. at maximal lateral pert.
Time [ms]	**885.9**	**890.2**	**892.8**	**899.4**	**912.6**	**921.5**	**924.2**
Event	Angle change onset of hip ABD at minimal lateral pert.	Peak MMG of hip ABD at medium medial pert.	Peak MMG of hip ABD atminimal medial pert.	Peak angle of hip ABD (5.5 ± 2.4 deg) at maximal medial pert.	Peak MMG of hip flex. at maximal medial pert.	Peak angle of hip ext. (1.8 ± 1.6 deg) at maximal medial pert.	Peak MMG of hip ext. at maximal lateral pert.
Time [ms]	**930.5**	**953.9**	**962.7**	**971.9**	**982.9**	**1016.2**	**1020.9**
Event	Peak power generating (0.05 ± 0.06 W/kg) at minimal lateral pert.	Hip ABD torque onset at minimal lateral pert.	Peak MMG of hip ext. at minimal medial pert.	Peak MMG of hip ext. at maximal medial pert.	Peak angle of hip flex. (6.9 ± 4.3 deg) at maximal lateral pert.	Peak angle of hip ADD (3.9 ± 2.3 deg) at maximal lateral pert.	Peak MMG of hip ext. at maximal medial pert.
Time [ms]	**1025.5**	**1026.4**	**1037.0**	**1039.3**	**1061.3**	**1066.7**	**1067.3**
Event	Angle change onset of hip ext. at medial lateral pert.	Hip ADD torque onset at minimal medial pert.	Peak of hip flex. torque (0.32 ± 0.13 Nm/kg) at maximal medial pert.	Peak MMG of hip ADD at maximal lateral pert.	Hip ABD torque onset at medial lateral pert.	Peak power absorbing (0.03 ± 0.02 W/kg) at minimal medial pert.	Hip ADD angle change onset of at minimal medial pert.
Time [ms]	**1069.8**	**1072.7**	**1092.8**	**1097.2**	**1122.8**	**1213.9**	**1215.1**
Event	Peak MMG of hip ADD at maximal medial pert.	Peak MMG of hip ABD at maximal medial pert.	Hip ABD torque onset at maximal lateral pert.	Peak MMG of hip ADD at maximal lateral pert.	Hip ext. torque onset at minimal medial pert.	Peak MMG of hip flex. at minimal medial pert.	Angle change onset of hip ext. at minimal medial pert.
Time [ms]	**1223.0**	**1228.2**	**1245.2**	**1251.8**	**1252.3**	**1256.7**	**1312.8**
Event	Peak angle of hip flex. (9.1 ± 5.9 deg) at maximal medial pert.	Peak angle of hip ext. (1.9 ± 1.3 deg) at maximal lateral pert.	Ext. torque onset at medium medial pert.	Peak power absorbing (0.04 ± 0.02 W/kg) at medium medial pert.	Peak of hip flex. torque (0.25 ± 0.15 Nm/kg) at maximal lateral pert.	Hip ext. torque onset at maximal medial pert.	Angle change onset of hip ext. at medium medial pert.
Time [ms]	**1318.1**	**1346.1**	**1347.1**	**1311.4**	**1352.7**	**1357.9**	**1360.2**
Event	Peak power absorbing (0.11 ± 0.09 W/kg) at maximal medial pert.	Peak of hip ABD torque (0.04 ± 0.03 Nm/kg) at minimal lateral pert.	Peak of hip flex. torque (0.15 ± 0.10 N m/kg) at medial lateral pert.	Peak angle of hip ABD (2.4 ± 1.4 deg) at maximal lateral pert.	Peak angle of hip ADD (2.8 ± 3.4 deg) at maximal medial pert.	Peak angle of hip ABD (1.1 ± 0.8 deg) at minimal lateral pert.	Peak angle of hip ABD (1.8 ± 0.7 deg) at medial lateral pert.
Time [ms]	**1370.8**	**1386.7**	**1387.2**	**1407.3**	**1417.8**	**1430.9**	**1432.0**
Event	Peak of hip ABD torque (0.04 ± 0.03 Nm/kg) at medial lateral pert.	Peak of hip ADD torque (0.11 ± 0.08 Nm/kg) at minimal medial pert.	Peak angle of hip ext. (1.1 ± 0.7 deg) at minimal lateral pert.	Peak angle of hip ext. (1.4 ± 0.4 deg) at medial lateral pert.	Peak of hip ADD torque (0.19 ± 0.10 Nm/kg) at medium medial pert	Peak of hip flex. torque (0.09 ± 0.05 Nm/kg) at minimal lateral pert.	Peak of hip ext. torque (0.09 ± 0.09 Nm/kg) at minimal medial pert.

Table A3. *Cont.*

Time [ms]	**1486.4**	**1586.0**	**1593.2**	**1613.1**	**1606.2**	**1665.6**	**1667.8**
Event	Peak angle of hip ADD (1.5 ± 0.5 deg) at medium medial pert.	Peak angle of hip ADD (1.1 ± 0.5 deg) at minimal medial pert.	Peak of hip ABD torque (0.08 ± 0.05 Nm/kg) at maximal lateral pert.	Peak angle of hip ext. (0.9 ± 0.7 deg) at minimal medial pert.	Peak of hip ADD torque (0.32 ± 0.19 Nm/kg) at maximal medial pert.	Peak of hip ext. torque (0.17 ± 0.11 Nm/kg) at maximal medial pert.	Peak of hip ext. torque (0.13 ± 0.10 Nm/kg) at medium medial pert.
Time [ms]	**1729.7**						
Event	Peak angle of hip ext. (1.4 ± 1.0 deg) at medium medial pert.						

Table A4. A detailed description of the temporal reactions to mediolateral perturbations of the CoP and CoM. The provided data are mainly from the paper [20].

Time [ms]	**86.7**	**93.1**	**94.6**	**95.2**	**96**	**100.5**	**101.1**
Event	CoP lateral displacement onset at maximal pert.	CoM medial displacement onset at maximal pert.	CoP medial displacement onset at medial pert.	CoP medial displacement onset at maximal pert.	Weight transfer onset at 50% preload [25]	CoP lateral displacement onset at medial pert.	CoM lateral displacement onset at maximal pert.
Time [ms]	**104**	**109.0**	**114.0**	**114.6**	**115.0**	**120**	**125.8**
Event	Weight transfer onseta at 65% preload [25]	CoM medial displacement onset at minimal pert.	CoM lateral displacement onset at medial pert.	CoP medial displacement onset at minimal pert.	CoP lateral displacement onset at minimal pert.	Weight transfer onset at 80% preload [25]	CoM medial displacement onset at medial pert.
Time [ms]	**131.2**	**140**	**150**	**156.6**	**191.1**	**234**	**622.4**
Event	CoM lateral displacement onsetat minimal pert.	GRF onset under the initial swing leg; LSS [5]	GRF onset under the initial swing leg; UCS [5]	CoP ML displacement onset [24]	CoP AP displacement onset [24]	Weight transfer time [19]	Peak CoP medial displacementat minimal pert. 86.7 ± 12.8 mm
				Left step, swing leg: CoP ML: 0.045 ± 0.009 m; CoP AP: −0.031 ± 0.006 m; stance leg: CoP ML: −0.074 ± 0.005 m; CoP AP: −0.051 ± 0.006 m			
				Right step, swing leg: CoP ML: −0.040 ± 0.007 m; CoP AP: −0.030 ± 0.004 m; stance leg: CoP ML: 0.089 ± 0.007 m; CoP AP: −0.049 ± 0.007 m			
Time [ms]	**642.7**	**656.6**	**662.4**	**674.7**	**721.1**	**746.8**	**756.7**
Event	Peak CoP lateral displacementat minimal pert. (74.2 ± 10.3 mm)	Peak CoP lateral displacement at medial pert. (145.3 ± 16.3 mm)	Peak CoP medial displacement at medial pert. (145.6 ± 11.7 mm)	Peak CoM medial displacementat minimal pert. (30.4 ± 10.4 mm)	Peak CoM lateral displacementat minimal pert. (30.3 ± 9.0 mm)	Peak CoM lateral displacement at medial pert. (60.7 ± 11.9 mm)	Peak CoM medial displacement at medial pert. (63.7 ± 12.1 mm)
Time [ms]	**798.9**	**835.2**	**991.3**	**975.6**			
Event	Peak CoP medial displacement at maximal pert. (174.2 ± 15.8 mm)	Peak CoP lateral displacement at maximal pert. (175.3 ± 13.4 mm)	Peak CoM lateral displacement at maximal pert. (119.1 ± 27.9 mm)	Peak CoM medial displacement at maximal pert. (122.2 ± 21.1 mm)			

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