# Postural Responses of Young Adults to Collision in Virtual World Combined With Horizontal Translation of Haptic Floor

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***Abstract*—Balance and postural response strategies change when subjects are exposed to horizontal translations of the floor or virtual reality or both. This may impact the balance training strategy and balance capabilities assessment in the future telerehabilitation. In the study 15 neurologically intact volunteers participated. Balance standing frame with virtual reality tasks and our novel haptic floor able to generate horizontal translations were used. The postural responses were measured with center of gravity and muscle electromyography of plantar-dorsiflexors, quadriceps, hamstrings, hip and spine muscles in three scenarios. The results demonstrated that center of gravity and electromyographic activity were comparable; with low latency at translation only, longer latency at combination with virtual reality and long latency when only virtual reality was applied. Soleus and semimembranousis demonstrated lower latency at back-right horizontal translations when virtual reality was present. The outcomes suggests that the postural strategy changes from ankle to ankle-hip strategy with availability of additional sensory systems which may be an important issue for objective balance evaluation in the clinical environment and remote telerehabilitation.**

***Index Terms*—Apparatus, balance training, berg balance scale, center of pressure, evaluation, fall, instrument, postural response, posture, rehabilitation, stroke, telerehabilitation, teletherapy.**

## I. Introduction

It IS estimated that nearly 15 million people around the world suffer for stroke each year [1]. According to the U.K. statistics the stroke occurred in one out of six people in the world, causing a death of nearly 1.1 million Europeans each year and may rise up to 1.5 million by the year 2025. The report reveals that stroke by itself is the second single most common cause of death in Europe/EU [1]. However, the consequences for 2/3 of the stroke survivors depends on how severe the stroke was and which parts of the brain were affected. Most of them partly recover after the intensive rehabilitation, they become functionally independent and achieve better quality of life. Intensive physiotherapy with a lot of repetitions and accuracy at tasks requires goal oriented tasks to achieve effective results [2]. Therefore, nowadays many rehabilitation centers use appropriate assisting devices to assure greater repeatability, better safety and task compliance. The assisting devices enable body weight support, trunk and pelvis stabilization and may also provide active support at standing, balancing, standing up or walking. Beside safe rehabilitation [3] it is important to overcome the subject's fear of falling as a possible consequence of poor balance confidence or deteriorated balance capabilities [4]. Active devices (e.g., KineAssist, Kinea Design, Evanston, IL, USA) comprise actuators, electro or other types of motors to actively support the subjects during upright posture, to actively control the desired posture or provide active support up to the level of subject's voluntary activity. On the other hand the passive devices usually limit the range of motion (e.g., balance) in selected joints or body, but require certain level of participant's voluntary activity. The advantages of both types of devices can take full effect when integrated with a target oriented task.

The task oriented rehabilitation involving the active role of the participants in the rehabilitation process demonstrated favorable outcomes in stroke population [5]. Thus the voluntary activity invokes changes in the primary motor cortex [6]. Ball exercises, reaching targets, weight transfer, etc., have been recently replaced by task designed in virtual reality (VR). The major advantages of VR over the real world task were found in attractiveness, motivation [7] and adaptability of the VR objects or difficulty levels according to the subjects' capabilities [8]. The latter is very important to keep up with the subjects' motivation, fatigue, boringness [9]. Considering the advantages of the target oriented VR tasks the authors reported on significant improvement of motor functions in treadmill walking [10], posture and balance [11]. VR enviroment already serving as a visual feedback could be successfully combined with haptic feedback; e.g., vibrotactile information on the floor. Such information may increase the action/reaction activities, help in identification of objects, floor and thus contribute to cognitive component in the rehabilitation [12]. Besides the aforementioned haptic feedback, the moving floor may also serve as a tool for generating platform perturbations. The platform perturbations elicit postural responses, actions that activate postural mechanisms [13] to prevent a fall and helps the subject to maintain upright posture without lifting or moving the feet. The postural mechanisms comprise ankle and hip strategies or combination of both, depending on the perturbation direction and strength. When the

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floor moves backward, it is very likely that the subjects would use the ankle strategy [14]. The differences in balance and postural strategies through the rehabilitation process have been objectively evaluated on translational or rotating platforms using kinematics, kinetics and electromyography (EMG) [16]–[18]. Keshner *et al.* [19] investigated postural kinematics in visually sensitive subjects during the combined postural perturbations and rotating visual field. They reported on disparate effects on stabilizing the subjects' trunk and head and concluded that VR technology may be used to identify sensitivity to visual motion. Further on Slaboda *et al.* [20] investigated the combination of vision and rotating platform in adults with cerebral palsy and reported that vision did not influence muscle frequency response in lower limb at stationary platform, but at the onset of platform tilt. However, to our knowledge combined VR balance training with haptic feedback has not been yet investigated in terms of postural responses when platform was translated during balance training. Recently we build a VR supported balance training (VRBT) system with haptic floor that enabled horizontal translation of the standing platform [21] at the collision with the VR object.

In this paper, we present a pilot study investigating the postural responses in healthy, neurologically intact adults standing in the balance frame for the three different situations; postural responses to randomly generated horizontal translations without visual information, VRBT without haptic feedback and a combination of VRBT and translational perturbation at the onset of collision in the VR world. Within all tasks monitoring of center of gravity and electromyography of major postural muscles were carried out. The presence of VR enabled the participants to prepare in advance for the postural perturbation. We hypothesized that subjects would choose the ankle complex to stabilize the calf as the platform moved backward with rather small perturbation magnitude and we expected that subjects' responses to horizontal translations will change when visual information in the VR world was present.

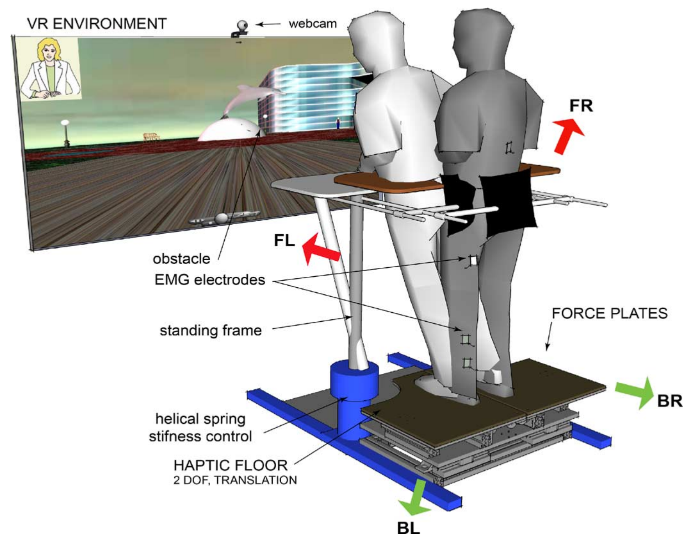


Fig. 1. In the study an existing telerehabilitation system for VR balance training with teleassistance and haptic floor was used. Three scenarios were applied: 1. blank screen with translational perturbation at the floor level; 2. constant speed movement in the virtual world and collision avoidance; 3. in addition to the 2. scenario the haptic floor translational feedback at the onset of collision with VR object. Haptic floor moved backward left—BL- (right—BR) at the collision with the virtual object, consequently the subjects leaned in the opposite direction—forward right (left), FR, FL, respectively. EMG of TA, SOL, GAS, RF, SEM, TFL, ESI muscles on the left lower extremity were measured in addition to the CoG assessment.

## II. Methods

### A. Equipment for Control and Data Collection

In the study a modified standing frame for balance training [22], now commercialized under the name BalanceTrainer (Medica Medizintechnik, Germany), was used in order to assure safety during standing. An upper frame was fixed on the base of the BalanceTrainer (BT) standing frame using passive controllable helical springs (Fig. 1). The BT's two degrees-of-freedom (2 DOF) tilt was mechanically limited within 15° in both sagittal and frontal plane and its stiffness was adjustable by a lever defining the length of the passive springs. Custom made three-axis tilt sensor (from Toradex, Switzerland components) was mounted on the frame and used to control the movement in the designed targeted task in VR environment. The information on avatar position, speed and collision was sent to the haptic floor controller via UDP/IP. The haptic floor [21] was designed to provide horizontal translations in all directions of the transversal plane and controlled in the way that could elicit postural responses. Aluminum profiles holding platforms in two layers and guide rails were used to enable smooth movement. DC motors (Maxon DC RE40, 150 W, Encoder HEDS 5540, Maxon, Planetary Gearhead GP 52, Switzerland) moved the lower layer using steel wires in medial-lateral (M/L) direction and upper platform in anterior-posterior (A/P) direction. TTL signals from encoders (A, B, and I) were sampled with National Instruments (NI) high-speed digital I/O module (NI 9403, USA). The real-time quadrature decoding and control algorithm were implemented in real-time controller (NI cRIO-9014, USA). The dc motors were controlled with analogue output (NI 9263 AO, USA) and the power for dc motors came from the servo amplifier Maxon 4QDC (ADS 50/10, pulsed (PWM) 4QDC Servo amplifier 50 V/10 A) The algorithms were written in Labview 8.5 FPGA (NI). Data synchronization between the haptic plate movement, VR avatar position, and external data assessment were established with trigger signal sent from NI controller to the digital input of the measurement equipment (Keithley KPCMCIA-12AI data acquisition card). The equipment recorded signals at sampling frequency of 1 kHz from haptic floor (HF) position, human kinematics and electromyographic (EMG) muscle activities. For EMG signal assessment Noraxon system (Noraxon 2000 EMG system, Noraxon Inc., Scottsdale, AZ, USA) with surface electrodes (3M Red Dot Repositionable Electrodes) was used. Custom made interface (MATLAB, The MathWorks, Inc., Natick, MA, USA) enabled a synchronization of the centre of gravity (CoG) (sampled at 200 Hz) assessed in force plates (made of Nintendo Wii Balance board sensors) with the measured data.

### *B. Subjects*

In the pilot study 15 neurologically intact young adults (11 male, four female 28.8 SD 2.9 years, 72.6 SD 15.6 kg, 173.8 SD 9.9 cm) participated voluntarily. The volunteers had no disease or muscular-skeletal or neurological impairment that may affect motor control, cognitive or vision capabilities. As the volunteers had no previous experience with the BT or with the designated VR task they were given 5 min short introduction including a short testing of the equipment and the task before the session.

The study was approved by local medical ethics committee and the subjects gave informed consent.

### *C. Assessment Protocol and Stimuli*

Subjects were standing in the standing frame and were asked to straighten their vision ahead, stand comfortably with feet in parallel, with each feet on the designated area on the force plate. Their arms were resting on the table that also moved together with the standing frame. The horizontal translations of the standing floor in eight different directions (forward: FW; backward: BW; left: LT; right: RT; forward-right: FR; forward-left: FL; backward-right: BR; and backward-left: BL) were considered as postural perturbations for the standing subjects. The perturbation intensity in neurologically intact subjects was set to the level that none of the subjects expressed any inconvenience. The perturbation magnitude (4 cm in 180 ms, peak acceleration 4.94 $\mathrm{m/s^2}$ with peak velocity 44.4 cm/s) was the same regardless of the angle of collision or weight of the subject due to the study conditions (Fig. 2). The subjects were instructed to stand quietly prior to the perturbation and immediately try to attain the upright posture when recovering from perturbation. The assessment protocol for identification of different postural strategies at VR presence and/or haptic information presence was divided into three scenarios.

1) The subjects stood in the standing frame quietly for 3–5 s then a perturbation took place. The perturbation onset (within 1 s) and the direction were randomly defined without prior notification of the subject. In approx. 2 s of time the platform slowly (approx. velocity 2.3 cm/s) returned to the initial position. The next perturbation occurred after 6–8 s. All eight directions of perturbation occurred within 80 s and the session was repeated three times. The recorded data were EMG, CoG, HF position, and direction of perturbation.
2) The subjects practiced balance in the VR environment [23] with predefined constrains. In this scenario the movement velocity was pre-set and remained constant through the complete session. The subjects controlled only the left-right movement in the medio-lateral (ML) plane by tilting the body and thus the standing frame and movement in the VR world. The important issue in the 80 s session (3–5 repetitions) was the collision with the object in the VR world during movement in forward direction. The subject had a clear vision of the VR object on the screen in front of him. The recorded data were EMG, CoG, position of the VR object and avatar and the direction of collision. No perturbation was present.

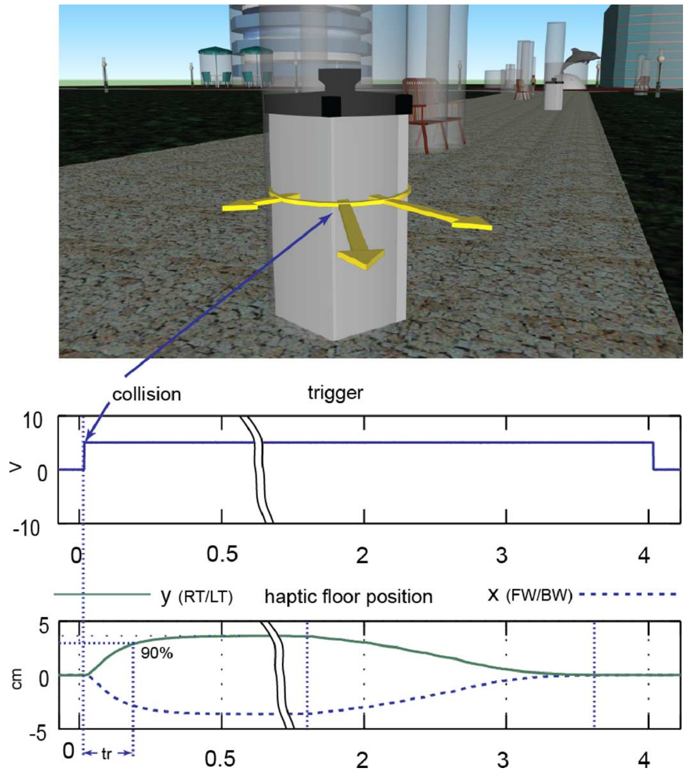


Fig. 2. At the onset of collision with VR object (upper figure) a trigger signal was generated to start the perturbation. Haptic floor horizontally translated with a short delay and reached the desired magnitude (4 cm) in $\mathrm{tr} = 180$ ms (peak acceleration 4.94 $\mathrm{m/s^2}$ with peak velocity 44.4 cm/s). After approx 1 s the platform slowly (2.3 cm/s) returned to the initial position.

3) The third assessment was identical to the second scenario with balance training in the VR environment [23] with predefined constrains, but with additional haptic feedback provided by horizontal translation of the HF. At the onset of collision with the VR object the HF moved in the opposite direction (BF, BW, or BR) according to the ideal collision law [21]. The result was a postural response of the subject standing on the HF platform. The system recorded EMG, CoG, position of the VR object and avatar, the direction of perturbation and the position of the HF.

The EMG activities in all three conditions were recorded on the left extremity and trunk; the left soleus (SOL), gastrocnemius (GAS), tibialis anterior (TA), semimembranosus (SEM), rectus femoris (RF), tensor fasciae latae (TFL), and erector spinae muscle at level L4 (ESI). However, it was assumed that EMG activity in healthy young adults was symmetrical [15].

### *D. EMG and COG Data Analysis*

At the absence of postural perturbation a weak EMG signal was expected considering constantly low muscle activity due to small body tilt and low game dynamics at VR supported balance training. The increased dynamics appeared just before and at the time of collision with the object in the VR world. Additionally the HF activity at the time of collision in both scenarios enormously increased the EMG activity. Therefore, the proposed bandpass filter 30–250 Hz [24] and full rectification of the signal was considered inappropriate as the rectification

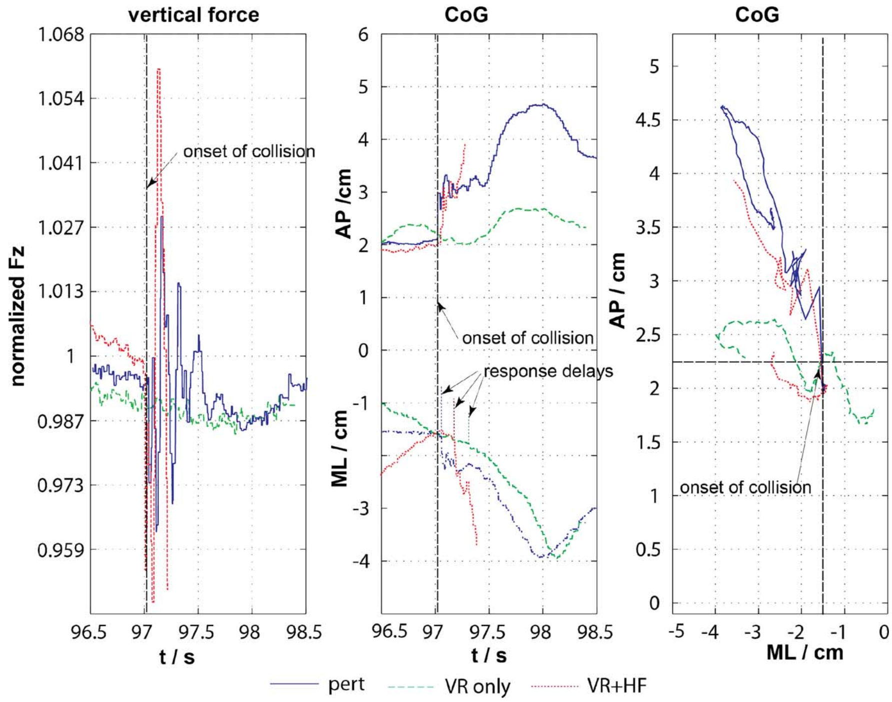


Fig. 3. Mean responses for all three conditions: perturbation only (pert), VR only, and VRHF. Vertical force (left) demonstrated rapid decrease and increase of the value normalized on the body weight for perturbation and VRHF conditions. Differences in response times between the three conditions are more clearly shown in the CoG track (middle). The sudden perturbation turned as the most reactive response, assuming that the VR allowed a certain preparation time. Desired transfer of CoG in the opposite direction than the perturbation was achieved only at the presence of HF perturbation without VR. Absence of HF movement resulted in a delayed (middle) and minimal CoG transfer in AP direction (right) and slow obstacle overcome by ML CoG transfer.

removed the sharp peaks in EMG data and thus disabled the identification of latencies. Instead of that recorded raw EMG data were low-pass filtered with 500 Hz Butterworth filter with bias removed and a continuous wavelet transformation (CWT) using the Morlet wavelet theoretically encompassing frequencies between 16 and 812 Hz was applied to show and detect the EMG activities at postural responses. CWT has proven as effective [20] for detecting abrupt transitions (MathWorks, Ltd.). More attention was paid to the bandwidth of 47–162 Hz, where the EMG activity was the strongest. Latencies for each measured EMG (TA, SOL, GAS, RF, SEM, TFL, ESI) were extracted manually from time course and CWT scalogram. The latencies were measured from the onset of HF movement or the impact with the object in the VR world to the first recorded muscle activity. Due to the partly weak S/N ratio, especially in EMG activity where no HF movement was present, the relevant time frame was accurately determined by scalogram, indicating the percentange of wavelet energy distribution across the scales. The scalogram at higher wavelet coefficients within the expected response was considered as the most likely onset of muscle response. Then the latencies were extracted from the raw EMG signal. As most of the collisions in the second and third scenario occurred in the diagonal directions, we decided to compare data assessed in BR and BL directions of perturbation.

EMG latencies at all three scenarios for BR and BL directions of perturbation were statistically examined (GNU PSPP 0.8.0 Freeware, MATLAB R2013–The MathWorks, Inc., Natick, MA, USA). The mean value and standard deviation (SD) were calculated from latencies obtained from each muscle EMG across the subjects. The values were compared among all three scenarios and both directions of perturbations (BR and BL) and differences between various conditions were statistically checked with repeated-measures ANOVA.

CoG movements in anterior–posterior (AP) and ML direction were calculated from data of four force sensors each in the corner of the force plate measuring the vertical component of the reaction force [25]; $\mathrm{x}$, $\mathrm{y}$ presented a fixed location of each sensor and $\mathrm{W}_i$ the measured data

$$\mathbf{AP} = \frac{\sum \mathbf{x}\mathbf{W}_i}{\sum \mathbf{W}_i}$$
$$\mathbf{ML} = \frac{\sum \mathbf{y}\mathbf{W}_i}{\sum \mathbf{W}_i}. \tag{1}$$

Data were sampled at 200 Hz and low-pass filtered with fourth-order Butterworth filter at 50 Hz. The common vertical force was calculated from four force sensor and normalized on each subject's weight.

## III. Results

### A. Center of Gravity at Postural Perturbation

The mean CoG responses at the onset of the collision or perturbation are shown in the Fig. 3 for all three scenarios. When the subjects were exposed to external perturbation in the BR direction by haptic floor only, an immediate response was noticed; in the normalized vertical force (Fig. 3 left) peaks with latency of 10–50 ms. Similar responses but with slightly longer latencies (20–70 ms) were noticed when VR was combined with HF

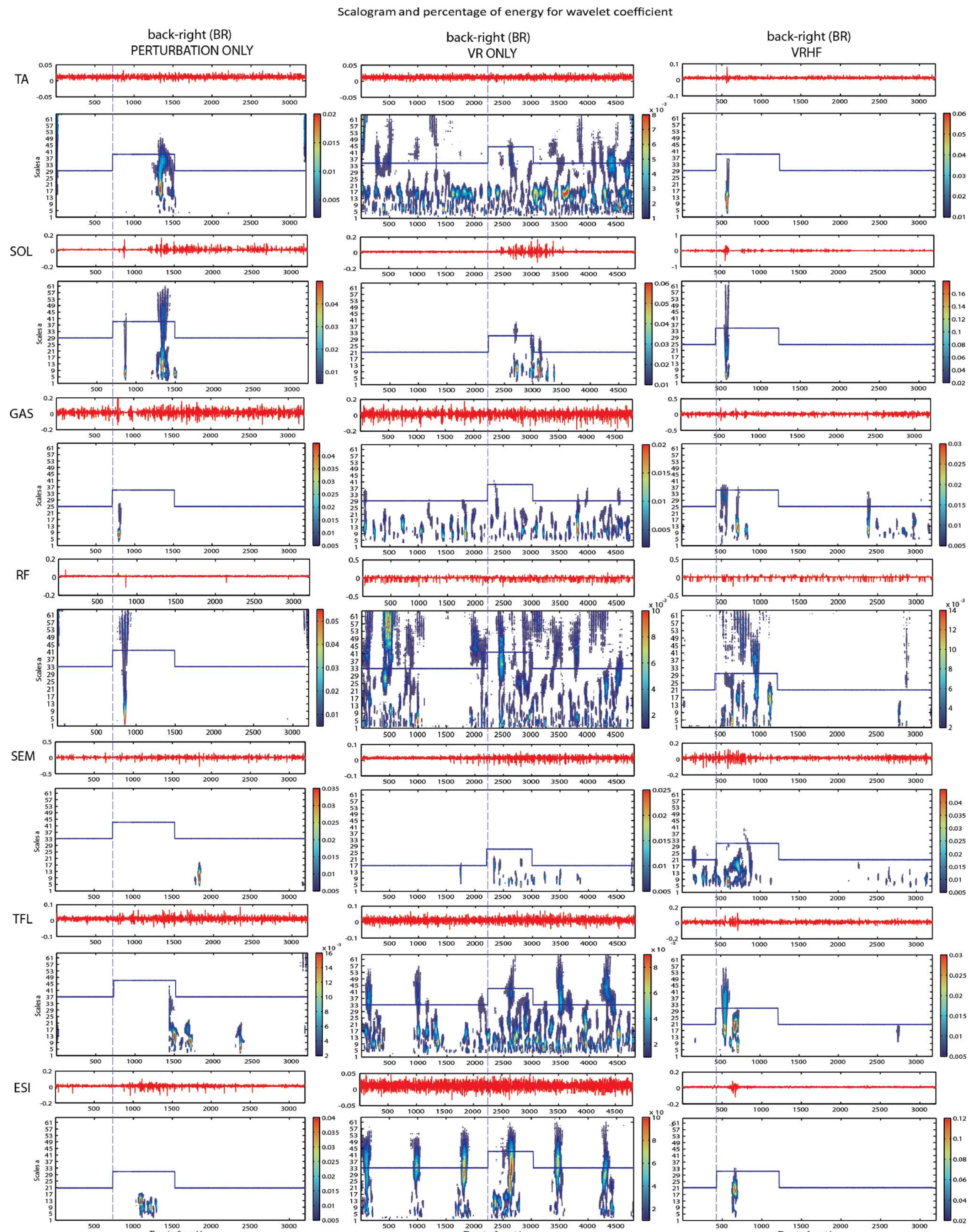


Fig. 4. EMG signals of the left TA, SOL, GAS, SEM, RF, TFL, and ESI at the onset of event (perturbation to the BR direction or collision in VR) for the representative subject are shown. Below the EMG signal is the scalogram with the percentage of energy for wavelet coefficients (dark—low percentage, bright higher percentage).

feedback. Hereby the subjects hit the obstacle on their FL and the HF responded with a perturbation in the BR direction.

The differences in response times between the three conditions are more clearly shown in the CoG time-course (Fig. 3 middle). In the upper part of the figure the AP response and in the lower part the ML responses are presented. The mean latency of CoG responses in AP direction were 41, 60, and 198 ms for HF perturbation scenario, VR including HF feedback (VRHF) and VR only scenario, respectively. In the ML direction the mean latency of CoG responses were 31, 155, and 279 ms, respectively.

The X-Y or AP-ML diagram (Fig. 3 right) demonstrates the spatial dispersion of the CoG responses in transversal plane. The perturbation induced an initial rapid response in the AP direction followed by a combination of the response in AP and ML direction.

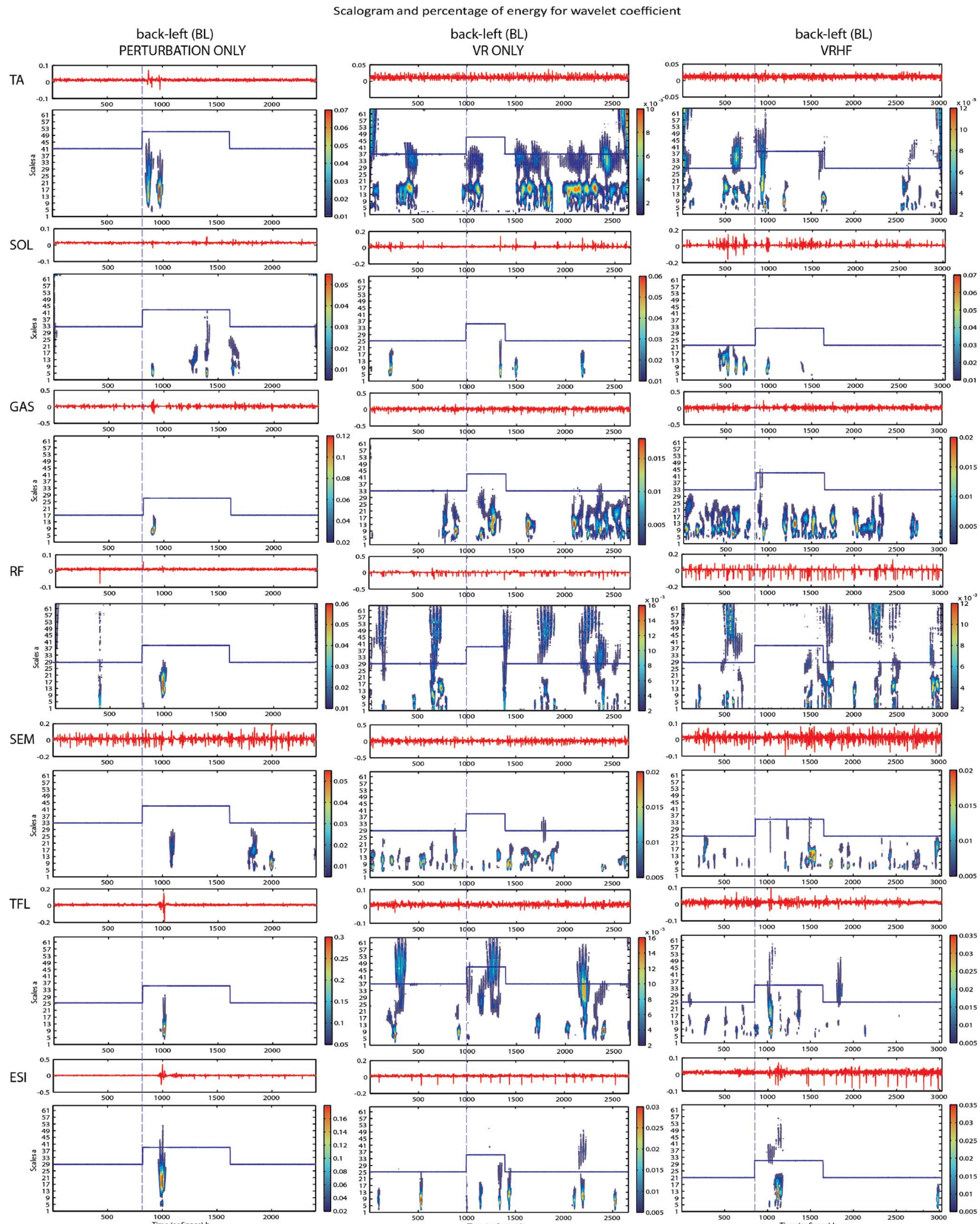


Fig. 5. Raw EMG signals of the left TA, SOL, GAS , SEM, RF, TFL, and ESI at the onset of event (perturbation to the BL direction or collision in VR) for the representative subject are shown. Bellow the scalogram shows how the raw EMG's energy was distributed across the scales.

## *B. Muscle Recruitment Latency*

The HF activity at the time of collision in both scenarios enormously increased the EMG activity, especially at combination of both, the VR feedback and HF perturbations, as seen in the scalogram of the representative subject in the right column of the Fig. 5. The left side distal muscles (TA, SOL, GAS) responded rapidly (100 ms) at BL translation of the HF, while proximal muscles (RF, SEM, TFL, and ESI) responded with larger latencies (150–200 ms). Larger RF latencies (320 ms) and premature activity of TA ($-120$ ms) were noticed at the scenario including also VR. In the scenario using VR only, muscles demonstrated much larger EMG latencies (300 ms). GAS and SEM muscles demonstrated premature activities ($-90$ ms, $-50$ ms) and (bright area in the middle column of Fig. 5) larger wavelet coefficients (0.07)—higher energy. RF was prematurely active ($-150$ ms), but the post response activity of the RF and TFL were rather low (low CWT coefficient

TABLE I
STATISTICAL DIFFERENCES IN EMG LATENCIES BETWEEN SCENARIOS

| | Back right | | | Back left |
|---|---|---|---|---|
| muscle | VR only vs pert | VR only vs VRHF | VRHF vs pert | VRHF vs pert |
| TA | 0.0006* | 0.0013* | 0.0537* | 0.3058 |
| SOL | 0.0778 | 0.0590* | 0.4366 | 0.9285 |
| GAS | 0.0002* | 0.0015* | 0.1096 | 0.2848 |
| RF | 0.0014* | 0.2302 | 0.2808 | 0.1719 |
| SEM | 0.0002* | 0.0001* | 0.1202 | 0.5172 |
| TFL | 0.0000* | 0.0000* | 0.7761 | 0.1858 |
| ESI | 0.0013* | 0.0041* | 0.9285 | 0.2021 |

*statistically significant

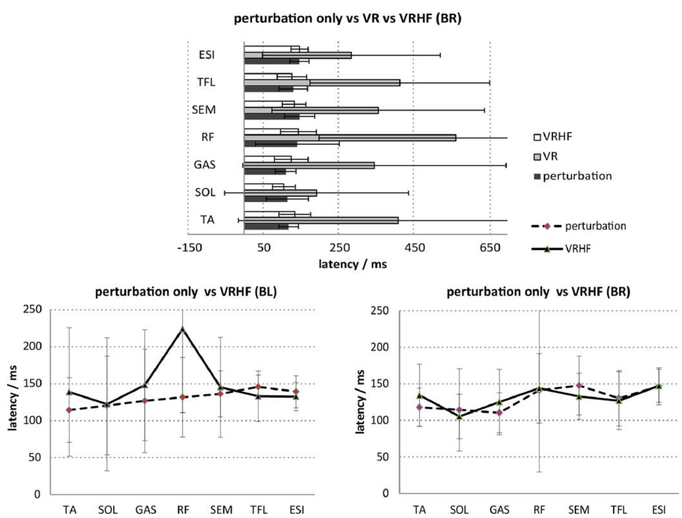


Fig. 6. Mean EMG latencies of the postural responses on back-right and back-left perturbations for all three scenarios; haptic floor translations (perturbation only), responses on visual input only (VR only) and combination of visual and HF perturbations at the onset of collision in the VR world. (VRHF). Bellow the comparisons of both scenarios comprising HF translations are shown.

at low frequencies). Similar differences were noticed in left side distal muscles EMG at HF translation in the BR direction. For the particular subject (Fig. 4) the TA muscle showed rather high latency (340 ms), but at scenario including also VR the TA latency was in line with other subjects (120 ms). The particular subject also demonstrated increased activities of TA, RF and TFL muscles in the scenario with VR only. The ESI muscle showed not only regular latency of 180 ms, but also repeated activity with rather high energy (CWT 0.01 at medium scales) in the scenario with VR only.

The upper Fig. 6 presents the mean values of EMG latencies for all participating subjects for all three scenarios. The quick comparison of EMG latencies for the BR translational perturbation shows that much longer latencies ($\mathrm{mean} > 200$ ms) were found in the responses to the VR obstacle than to the actual translational perturbation ($\mathrm{mean} < 150$ ms). For the particular direction of VR impact (BR) the most rapid response was noticed in the SOL muscle (192 ms), but with large SD (244 ms) and the RF (562 ms) response demonstrated the largest latency. Mean EMG latencies at perturbation in the BR direction were much shorter, GAS, TA, and SOL being the fastest, 110, 117, and 114 ms, respectively. The latencies in SEM, ESI, and RF muscles responses were a larger, 147, 146, and 141 ms, respectively. In the third scenario, where the perturbation in the BR direction was the consequence of VR collision (VRHF), the EMG latencies of SOL (105 ms), SEM (132 ms), and TFL (126 ms) were shorter than at perturbation only, but the TA (134 ms) and GAS (125 ms) were larger. There were practically no change in RF (144 ms) and ESI (147 ms) latencies. However, only the difference in the TA responses was marginally significant ($p = 0.0537$). All differences were considered significant ($p < 0.05$), when comparing latencies with VR only scenario, but SOL and RF (Table I). Comparison of perturbation only and perturbation including VR in the BL direction (lower left Fig. 6) demonstrated larger latencies of TA (138 ms versus 114 ms), SOL (122 ms versus 120 ms), GAS (148 ms versus 126 ms), RF (224 ms versus 126 ms), SEM (145 ms versus 136 ms), but shorter latencies of TFL (133 ms versus 145 ms), and ESI (132 ms versus 139 ms). However, changes in the BL direction were not statistically significant (Table I).

## IV. Discussion

Maintaining balance in upright posture is certainly one of the most demanding tasks that must be performed with neuromuscular system. The central nervous system perceive information mainly from three major sensory systems; vision, vestibular system and somatosensory system [17]. Vestibular system acquires linear and angular accelerations of our body. The somatosensory system as a multisensory system in our body senses the velocity, position, and orientation to gravity of body segments and their contact with outer world. Vision is our primary system for path planning and obstacle avoidance. Our results demonstrated how these systems have been actively involved in postural control and how the postural response strategy may change if one or more of these systems are excluded.

In the first scenario the subjects were exposed to perturbations by translational surface, so the major active sensory systems applied were somatosensory and also vestibular systems. In quiet standing the stabilizing mechanism is based on ankle strategy controlling the CoG in the AP plane and hip strategy the ML plane [17], but in our case with surface translation as a combination of both directions (BR) the early activation of proximal muscles was required. Our results showed early activations with low latencies in left GAS (also SOL), but later TFL, SEM and after that ESI, which was reported characteristical for backward translation [17]. However, a combination of backward and right translation also activated TA and RF and TFL in the same order. The CoG results confirm earlier postural response in anterior direction and a moment later to the lateral direction. The early activation of TFL may play an important role as a prime muscle to exert the hip torque and thus in combination with RF and SEM control the reaction in diagonal direction also in our study. However, in the BL direction the similar activation order of right lower extremity muscles can be expected [26] as we may assume that EMG activity in healthy subjects was symmetrical [15]. The latencies were in line with the findings and explanation of Hughes *et al.* [16].

The presence of VR enabled visual feedback and thus allowed the subject sufficient time to prepare for the impact. Consequently the postural responses were smoother and followed the expected direction. At the onset of perturbation (and VR collision) in the BR direction the CoG at first followed the AP plane and then the ML plane, but keeping the FL diagonal direction. The timing of muscle recruitment was longer in TA and GAS, but latency was shorter in SOL and SEM muscles meaning that more hip strategy was used. The explanation may be that the task requiring both somatosensory and visual information may be destabilizing. Findings in healthy adults that has been presented a visual information but with altered somatosensory information demonstrated changed balance activity [20]. At the unrealiable somatosensory information the visual feedback may cause delays in both, AP and ML directions. People with neuromuscular impairment usually rely more on visual input than somatosensory or vestibular system at keeping balance, especially children with cerebral palsy [27]. At perturbation in the BL direction the left (contralateral to the movement) lower extremity muscles demonstrated more hip strategy, with shorter TFL and ESI latencies, but much longer RF latency.

Unlike in the both conditions comprising HF translations, the responses at scenario with visual feedback only demonstrated predominantly ML movement (CoG) in order to avoid or overcome the collision in the VR world. The timing of left lower extremity and trunk muscles substantially differed from the scenarios where the feedback information mostly relied on somatosensory or vestibular system. The power of EMG signal was rather weak at the absence of postural perturbation due to the small body tilt and low game dynamics. The increased dynamics appeared just before and at the time of collision with the object in the VR world. However, the visual feedback in a way presented a warning for the participating subject that often demonstrated premature muscle activities before impact or long EMG latencies due to the lack of mechanical perturbation.

All in all the EMG latencies for TA, GAS, RF, and ESI in the BR direction of perturbation for all three scenarios were in line with the outcomes of CoG. Both, the EMGs in those muscles and CoG responses for AP and ML direction, demonstrated shortest latencies for the perturbation only scenario. When VR was added, the latencies were reported longer in EMG and CoG. However, in the VR only scenario both assessed values in EMG and CoG demonstrated significantly larger latencies. Similar findings were reported by Müller *et al.* [24] for plantar flexors and TA versus vertical force and CoG. This lead us to thought that force platforms or perhaps simple CoG assessment may replace the time consuming EMG assessment in everyday clinical practice, especially when considering our findings [28] and findings of Stevenson *et al.* [29] who reported that those objective measurements correlated with clinical evaluation of balance in persons with chronic hemipareses after stroke. Besides, clinical tools for functional measures are not capable of differentiating between the physiologic recovery and compensatory strategies [30] nor predicting directionality. Also a simple balance training using VR technology only can not provide a sufficient insight into postural response strategies due to the low dynamics. The findings of the study demonstrated that carefully designed targeted training of postural responses may contribute to the change of balance strategies in persons suffering for stroke and thus improve their dynamic balance capabilities and reduces the risk of falls. However, a combination of haptic floor and VR with appropriate CoG based balance capabilities evaluation tool may present a novelty in rehabilitation medicine and may significantly influence on telerehabilitation and telediagnostic services in patients [23].

We have noticed that relying on visual input only provided at the collision onset required a delayed response, but the combination of VR and perturbation demonstrated that subjects choose the strategy even before the onset. Studies [13] also reported that healthy subject's appear to preselect their stepping limb to reduce the instability when the perturbation characteristics are unpredictable.

## V. Conclusion

In the study, we have shown that not only ankle strategy was applied at horizontal translations, but when the named haptic floor was combined with VR also a hip strategy was used. However, the VR provided additional visual input information to the subject to prepare for unpredicted perturbation. This changed the compensatory strategy to the level that hip mono-articular and bi-articular muscles may be more synchronized with distal muscles in the combined compensatory strategy in order to avoid unpredictable instability.

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