KATEDRA BIOMECHATRONIKI

WYDZIAŁ INŻYNIERII BIOMEDYCZNEJ

Politechnika Śląska

Ocena zmian wybranych mechanizmów kontroli posturalnej w odpowiedzi na rzeczywiste i wirtualne bodźce prowadzące do wytrącenia z równowagi

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Na początku chciałabym szczególnie podziękować moim promotorom, dr. hab. inż. Jackowi Jurkojciowi, prof. PŚ oraz dr. inż. Piotrowi Wodarskiemu za nieocenioną pomoc, merytoryczne wsparcie oraz zaufanie okazane podczas pracy nad rozprawą.

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Dziękuję także wszystkim uczestnikom badań oraz tym, którzy w jakikolwiek sposób przyczynili się do powstania tej pracy.

Spis treści

Streszczenie

Zdolność utrzymywania równowagi ciała przez człowieka jest kluczowym elementem codziennego funkcjonowania człowieka, wpływającym na samodzielność i jakość życia. Utrzymywanie równowagi to dynamiczny proces, w którym ważną rolę odgrywają mechanizmy przygotowania posturalnego. Mechanizmy, takie jak wczesne, antycypacyjne oraz kompensacyjne przygotowanie posturalne, odpowiadają za przygotowanie i reakcje ciała na destabilizujące bodźce. Metody badania równowagi opierają się głównie na analizie przemieszczeń środka nacisku stóp w dziedzinie czasu i częstotliwości. Nowoczesne narzędzia, takie jak czujniki IMU oraz wirtualna rzeczywistość, oferują nowe możliwości badawcze, szczególnie w kontekście badań w sytuacji konfliktu sensorycznego i diagnostyki neurologicznej, w tym choroby Parkinsona. Analizy te są jednak niewystarczające, ponieważ pomijają szybkie, niecykliczne zmiany. które mogą być również bardzo istotne w diagnozowaniu i leczeniu pacjentów z zaburzeniami neurologicznymi, ortopedycznymi lub przedsionkowymi.

Zaobserwowana potrzeba rozwoju metod pomiarowych oraz nowych sposobów analizy danych w celu lepszego zrozumienia mechanizmów kontroli równowagi u człowieka doprowadziła do sformułowania trzech głównych celów badawczych: opracowanie metodyki pomiarów, które umożliwiają ocenę zmian w strategii kontroli posturalnej w odpowiedzi na spodziewane i niespodziewane bodźce destabilizujące, określenie wpływu bodźców wirtualnych i rzeczywistych na przygotowanie posturalne jako narzędzie diagnostyczne oraz analiza możliwości praktycznego zastosowania metody wykrywania chwilowych korekt postawy w ocenie strategii kontroli posturalnej.

Rozprawa doktorska stanowi podsumowanie wyników badań opublikowanych w szeregu artykułów naukowych, które koncentrują się na mechanizmach kontroli posturalnej i zdolności utrzymywania równowagi u człowieka. Zbiór ten obejmuje zagadnienia dotyczące zarówno teoretycznych aspektów kontroli postawy, w tym opracowanej nowej metodyki analizy danych stabilograficznych, jak i praktycznych zastosowań klinicznych. W opisywanych artykułach stopniowo przechodzono od

prostszych eksperymentów oceniających zdolność utrzymywania równowagi ciała w wirtualnej rzeczywistości, przez wprowadzenie bardziej złożonych bodźców destabilizujących w postaci przesunięcia podłoża w świecie rzeczywistym, aż po rozwój nowych metod analizy danych opartych o wskaźniki stosowane na giełdzie, które mają na celu uzupełnienie tradycyjnych technik oceny równowagi, kończąc na praktycznym wykorzystaniu opracowanych metod w praktyce klinicznej na grupie osób z chorobą Parkinsona. Wykazano, że standardowe metody oceny zdolności utrzymywania równowagi, takie jak analiza wielkości w dziedzinie czasu i częstotliwości, okazały się niewystarczające do pełnego określenia reakcji na bodźce. Zastosowanie badań z bodźcami destabilizującymi, zarówno rzeczywistymi jak i wygenerowanymi w technologii wirtualnej rzeczywistości znacząco poszerza możliwości analizy zdolności utrzymywania równowagi. Wykorzystanie metod detekcji chwilowych korekt postawy, takich jak analiza zmian trendu w sygnale COP, zwiększa możliwości interpretacyjne zjawisk towarzyszących destabilizacji ciała osoby badanej.

Summary

The ability to maintain balance is a crucial element of daily human functioning, influencing both independence and quality of life. Maintaining balance is dynamic, with postural preparation mechanisms playing a significant role. Mechanisms such as early, anticipatory, and compensatory postural adjustments prepare the body and respond to destabilizing stimuli. Methods for studying balance primarily focus on analyzing center of pressure (COP) displacements in both time and frequency domains. Modern tools, such as IMU sensors and virtual reality, offer new research opportunities, particularly in the context of sensory conflicts and neurological diagnostics, including Parkinson's disease. However, these analyses can be insufficient, as they often overlook rapid, non-cyclical changes that may be critical for diagnosing and treating patients with neurological, orthopedic, or vestibular disorders.

The observed need for developing measurement methods and new approaches to data analysis for a better understanding of balance control mechanisms in humans led to the formulation of three main research objectives: (1) to develop measurement methodologies that enable the assessment of changes in postural control strategies in response to expected and unexpected destabilizing stimuli, (2) to determine the impact of virtual and real stimuli on postural preparation as a diagnostic tool, and (3) to analyze the practical application of methods for detecting momentary postural corrections in assessing postural control strategies.

The doctoral dissertation summarizes research findings published in several scientific articles, which focus on postural control mechanisms and the ability to maintain balance in humans. This body of work addresses both theoretical aspects of postural control, including a newly developed methodology for analyzing stabilographic data, and practical clinical applications. The articles gradually progressed from simpler experiments assessing balance ability in virtual reality to more complex destabilizing stimuli, such as real-world surface shifts, and to the development of new data analysis methods based on stock market indicators. These new methods aim to complement traditional balance assessment techniques and

culminated in the clinical application of the developed methods in a group of individuals with Parkinson's disease. It was shown that standard methods of assessing balance ability, such as time- and frequency-domain analyses, were insufficient for fully capturing reactions to stimuli. The use of destabilizing stimuli, both real and generated through virtual reality, significantly expanded the possibilities for analyzing balance abilities. Methods for detecting momentary postural corrections, such as trend analysis in the COP signal, enhanced the interpretive capacity for understanding phenomena associated with body destabilization.

1. Wykaz publikacji stanowiących rozprawę doktorską

Niniejsza praca składa się ze zbioru opublikowanych i powiązanych tematycznie artykułów:

[A1] Wodarski Piotr, Jurkojć Jacek, **Chmura Marta**, Gruszka Grzegorz, Gzik Marek. Analysis of center of pressure displacements and head movements triggered by a visual stimulus created using the virtual reality technology. *Acta Bioeng Biomech*. 2022;24(1):19-28. DOI: 10.37190/ABB-01900-2021-01. *Punktacja MNiSW = 100. IF $^{(2022)}$ = 1,000.*

[A2] Wodarski Piotr, **Chmura Marta**, Szlęzak Michał, Gruszka Grzegorz, Romanek Justyna, Jurkojć Jacek. The effect of selected lower limb muscle activities on a level of imbalance in reaction on anterior-posterior ground perturbation. *Acta Bioeng Biomech*. 2022;24(3):135-146. DOI: 10.37190/ABB-02112-2022-02.
Punktacja MNiSW = 100. IF $^{(2022)}$ = 1,000.

[A3] Wodarski Piotr, **Chmura Marta**, Gruszka Grzegorz, Romanek Justyna, Jurkojć Jacek. The stock market indexes in research on human balance. *Acta Bioeng Biomech*. 2022;24(2):163-176. DOI: 10.37190/ABB-02062-2022-03. *Punktacja MNiSW = 100. IF $^{(2022)}$ = 1,000.*

[A4] Wodarski Piotr, **Chmura Marta**, Jurkojć Jacek. Impact of Visual Disturbances on the Trend Changes of COP Displacement Courses Using Stock Exchange Indices. *Applied Sciences*. 2024; 14(11):4953. DOI: 10.3390/app14114953. Punktacja MNiSW = 100. IF $^{(2023)}$ = 2,500.

[A5] Wodarski Piotr, **Chmura Marta**, Szlęzak Michał, Bajor Grzegorz, Gzik Marek, Jurkojć Jacek. Trend change analysis in the assessment of body balance during posture adjustment in reaction to anterior-posterior ground perturbation. *PLoS One*. 2024;19(4): e0301227. DOI: 10.1371/journal.pone.0301227.
Punktacja MNiSW = 100. IF $^{(2023)}$ = 2,900.

[A6] Wodarski Piotr, Jurkojć Jacek, **Chmura Marta**, Elke Warmerdam, Robbin Romijnders, Markus A Hobert, Walter Maetzler, Krzysztof Cygoń, Clint

Hansen. Trend change analysis of postural balance in Parkinson's disease discriminates between medication state. *J Neuroeng Rehabil*. 2024;21(1):112. Published 2024 Jun 28. DOI: 10.1186/s12984-024-01411-z.

Punktacja MNiSW = 140. IF $^{(2023)}$ = 5,200.

2. Poszerzone streszczenie w języku polskim

2.1 Wprowadzenie

Zdolność utrzymywania równowagi jest kluczowym elementem codziennego funkcjonowania człowieka. Wpływa ona na niezależność w wykonywaniu czynności dnia codziennego, możliwość samodzielnego poruszania się czy też szeroko pojętą jakość życia. Utrzymywanie równowagi to złożony proces angażujący między innymi układ przedsionkowy układ wzrokowy oraz układ proprioceptywny. Układ przedsionkowy Jest zlokalizowany w uchu wewnętrznym i składa się z kanałów półkolistych oraz woreczka i łagiewki, które reagują na przyspieszenie głowy [1]. Zmysł wzroku dostarcza informacji o otaczającym świecie, takich jak natężenie światła, położenie obiektów w zasięgu wzroku czy położenie widocznych części ciała. Układ proprioceptywny dostarcza sensorycznych informacji o wzajemnym położeniu części ciała względem siebie. Zakończenia czuciowe proprioceptorów są wrażliwe na deformację. Istnieje kilka grup proprioceptorów, które są szczególnie ważne dla kontroli ruchu. Są one wrażliwe na zmienne fizyczne, takie jak pozycja stawu, długość prędkość skurczu i siła mięśnia. Integracja informacji z wcześniej wymienionych układów jest niezbędna do utrzymania równowagi. Zaburzenia któregokolwiek z tych układów mogą prowadzić do problemów ze zdolnością utrzymywania równowagi i zwiększonego ryzyka upadków. Zaburzenia w zdolności utrzymywania równowagi mogą być wynikiem chorób układu nerwowego, procesu starzenia się, przedsionkowych zaburzeń równowagi, zaburzeń w czuciu w stopach w wyniku cukrzycy lub obrażeń narządu ruchu [1,2]. Zaburzona zdolność utrzymywania równowagi może prowadzić do upadków, które skutkując urazami, mogą dodatkowo pogorszyć stan zdrowia [3]. Upadki oraz powstałe w ich wyniku obrażenia są jedną z głównych przyczyn hospitalizacji osób starszych, co w efekcie może prowadzić do utraty mobilności oraz utraty samodzielności życiowej [4]. W związku z tym, kluczowe dla poprawy jakości życia, szczególnie w populacji starszej oraz wśród osób z chorobami przewlekłymi, jest monitorowanie i poprawa zdolności utrzymywania równowagi, a przez to ograniczenie ryzyka upadków.

Utrzymanie stabilnej postawy ciała jest dynamicznym procesem wymagającym ciągłej kontroli oraz korekt pozycji środka masy w celu utrzymania równowagi podczas wykonywania czynności życia codziennego. Proces utrzymywania równowagi opiera się na procesach związanych z przygotowaniem posturalnym (PA) oraz procesem kompensacji posturalnej w odpowiedzi na bodziec destabilizujący. Przygotowanie posturalne występuje tuż przed zaburzeniem posturalnym, a jego rolą jest wyeliminowanie lub minimalizacja negatywnych skutków utraty równowagi. Natomiast korekty kompensacyjne mają na celu przywrócenie równowagi ciała tuż po wystąpieniu zaburzenia [5, 6, 7]. Mechanizmy kontroli postawy mogą być wczesne (ang. early postural adjustment – EPA) – występujące ok 600-400 ms przed zaburzeniem, przygotowujące ciało na nadchodzące zaburzenie, antycypacyjne (ang. anticipatory postural adjustment – APA) – występujące ok 150 ms przed do 50 ms po wystąpieniu zaburzenia, przygotowujące do przeciwstawienia się zaburzeniu, lub kompensacyjne (ang. compensatory postural adjustment – CPA) występujące ok 70-300ms po wystąpieniu zaburzenia, przywracające równowagę po zaburzeniu [8, 9, 10]. Występowanie wyżej opisanych mechanizmów objawia się głównie zmianą aktywności mięśniowej mięśni posturalnych [8], przemieszczeniem poszczególnych segmentów ciała, przemieszczeniem środka masy (COM) lub środka nacisku stóp (COP) [6]. Bodźce wywołujące PA można podzielić na dwa typy – pierwszy, związany z dobrowolnym zainicjowaniem ruchu [11] oraz drugi, zewnętrzny, pochodzący z otaczającego środowiska, często prowadzący do destabilizacji [5]. Badania sugerują, że bodziec zewnętrzny może wywołać APA, pod warunkiem, że osoba badana wie, kiedy wystąpi zaburzenie [8, 10]. W kontekście rehabilitacji i terapii, zrozumienie tych mechanizmów jest kluczowe, ze względu na potrzebę identyfikacji specyficznych deficytów w systemie kontroli równowagi pacjenta [12, 13], redukcji ryzyka upadków [4, 14] i monitorowania postępów rehabilitacji [15].

Naukowcy korzystają z różnorodnych sposobów, które umożliwiają ocenę zdolności utrzymywania równowagi. Najczęściej stosowana jest metoda bazująca na analizie przemieszczeń środka nacisku stóp. Wielkości opisujące zdolność utrzymywania stabilnej postawy ciała na podstawie przemieszczeń COP najczęściej analizowane są w dziedzinie czasu [16, 17] i w dziedzinie częstotliwości [18, 19]. Do najpopularniejszych wielkości w dziedzinie czasu należą: długość ścieżki COP, średnia prędkość COP, pole powierzchni elipsy COP oraz zakresy ruchu COP w kierunkach przednio-tylnym (AP) i środkowo-bocznym (ML). Testy mogą również obejmować określenie pozycji COM, ruchów głowy oraz wartości przyspieszenia tułowia [20]. W literaturze opisywane są także dużo rzadziej używane metody, takie jak analiza z wykorzystaniem rozkładu falkowego [21, 22, 23] czy też analizy probabilistyczne oparte na entropii [24].

Analizy w dziedzinie częstotliwości rozszerzają analizy w dziedzinie czasu o identyfikację składowych cyklicznych, które pojawiają się w sygnale COP. Analizy częstotliwościowe mogą ujawnić subtelne zmiany, które są trudno zauważalne w standardowych analizach opartych na wielkościach w dziedzinie czasu. Mogą one odgrywać rolę w opracowywaniu nowych metod diagnostyki w przypadku analizy zdolności utrzymywania równowagi w sytuacji konfliktu sensorycznego – zarówno w świecie rzeczywistym jak i przy wykorzystaniu wirtualnej rzeczywistości. Technologia wirtualnej rzeczywistości w ostatnich latach zyskuje na popularności, a dzięki rozwojowi techniki jest coraz szerzej dostępna. Wykorzystanie trójwymiarowych obrazów pozwala na wygenerowanie środowiska i bodźców, które byłyby trudne do wytworzenia w rzeczywistym świecie. Dodatkowo, szerokie możliwości przygotowania scenariuszy umożliwiają dostosowanie procesu diagnostycznego i rehabilitacyjnego do konkretnego przypadku i umiejętności pacjenta. Sprawia to, że technologia wirtualnej rzeczywistości jest coraz częściej wykorzystywana w rehabilitacji czy diagnostyce, w tym schorzeń związanych z osłabioną zdolnością utrzymywania równowagi. Szczególnie przydatne w tym przypadku wydają się badania w sytuacji konfliktu bodźców sensorycznych. W takiej sytuacji poszczególne zmysły otrzymują sprzeczne informacje, a osoba badana jest poddawana bodźcom, które bezpośrednio lub pośrednio destabilizują jej postawę. Przykładem może być badanie, podczas którego osoba badana stoi stabilnie na nieruchomym podłożu, a za pomocą technologii wirtualnej rzeczywistości przekazywana jest do zmysłu wzroku informacja o ruchu otoczenia (np. poprzez wyświetlanie oscylującego pomieszczenie) [18, 25, 26, 27]. Analiza reakcji ciała na zaburzenia wizualne może być cennym źródłem informacji diagnostycznych dotyczących zdolności utrzymywania równowagi i adaptacji w dynamicznie zmieniających się warunkach otoczenia. Dysfunkcje w tej adaptacji mogą być wczesnym wskaźnikiem problemów neurologicznych, takich jak choroba Parkinsona czy zaburzenia przedsionkowe [28].

Stosunkowo nowym trendem w badaniach zdolności utrzymywania równowagi są pomiary z wykorzystaniem czujników bezwładnościowych wyposażonych w akcelerometr, żyroskop i magnetometr (IMU). Ich zdolność do dokładnego pomiaru kinematyki ciała w trzech wymiarach pozwala na bardziej kompleksową ocenę równowagi [29, 30, 31, 32]. Analiza danych z czujników IMU może dostarczyć dodatkowych informacji, na temat mechanizmów utrzymania równowagi i reakcji na zaburzenia, takich jaki zmiana kątów w stawach czy przyspieszenia liniowego tułowia [28, 31, 32].

Obecnie stosowane metody oceny zdolności utrzymywania równowagi, takie jak analiza przemieszczeń COP w dziedzinie czasu [16, 17] i częstotliwości [18, 19], dostarczają cennych informacji o stabilności posturalnej. Do powszechnie analizowanych parametrów w dziedzinie czasu opisujących zdolność do utrzymania równowagi należą prędkość COP, powierzchnia elipsy COP oraz zakresy ruchu COP w kierunkach AP i ML. Podczas analizy wcześniej wymienionych wartości, zakłada się, że wzrost wartości tych wielkości najczęściej wskazuje na występujące problemy z utrzymywaniem równowagi [33]. Analizy częstotliwościowe pozwalają na analizę cyklicznych komponentów mierzonego sygnału, identyfikując występujące w nim dominujące częstotliwości, co z kolei umożliwia ocenę pojawiających się zmian w strategii utrzymywania równowagi ciała przez człowieka. Informacje te są szczególnie istotne w przypadku badania oddziaływania cyklicznych zaburzeń w wirtualnej rzeczywistości [21, 22, 23] i rozszerzają możliwość oceny zdolności utrzymywania równowagi w stosunku do analiz w dziedzinie czasu. Wprowadzenie bardziej zaawansowanych technologii, jak badania z wykorzystaniem czujników IMU, elektromiografii oraz technologii wirtualnej rzeczywistości może wspomóc ocenę zdolności utrzymywania równowagi i przygotowania posturalnego [29, 30, 31, 32], wymaga jednak stosowania analiz, które pozwolą na pełne poznanie i interpretację informacji zawartych w sygnale pomiarowym. Opisane wyżej metody pozwalają na wykrycie w rejestrowanym sygnale zmian wskazujących na pojawiające się zmiany w strategii utrzymywania równowagi. Analizy te bazują jednak przede wszystkim na globalnej analizie sygnału lub wykorzystują cykliczny charakter pojawiających się zjawisk, nie uwzględniając występujących szybkich zmian o charakterze niecyklicznym, które mogą być również bardzo istotne w diagnozowaniu i leczeniu pacjentów z zaburzeniami neurologicznymi, ortopedycznymi lub przedsionkowymi. Istnieje więc potrzeba ciągłego rozwoju wykorzystywanych metod badawczych, które uzupełniając obecnie stosowane pomiary i analizy umożliwią pełniejsze poznanie i interpretację zmian pojawiających się w zdolności utrzymywania równowagi będących wynikiem rozwoju wielu chorób.

2.2 Cel pracy

Zaobserwowana potrzeba rozwoju metod pomiarowych oraz poszukiwania nowych metod prowadzenia analizy danych pomiarowych zapewniających lepsze zrozumienie mechanizmów kontroli utrzymywania równowagi ciała przez człowieka pozwoliła na sformułowanie następujących celów badawczych:

- Opracowanie metodyki prowadzenia pomiarów zdolności utrzymywania równowagi ciała przez człowieka umożliwiających ocenę zmian w strategii kontroli posturalnej jako reakcji na spodziewane i niespodziewane bodźce wytrącające z równowagi.
- Określenie wpływu wirtualnych i rzeczywistych bodźców wytrącających z równowagi na zjawisko przygotowania posturalnego jako narzędzia do diagnostyki zdolności utrzymywania równowagi ciała przez człowieka.
- Analiza możliwości praktycznego zastosowania metody detekcji chwilowych korekt postawy do oceny zmian w strategii kontroli posturalnej w odpowiedzi na wirtualne i rzeczywiste bodźce wytrącające z równowagi.

Niniejsza rozprawa doktorska stanowi podsumowanie wyników badań opublikowanych w szeregu artykułów naukowych, które koncentrują się na mechanizmach kontroli posturalnej i zdolności utrzymywania równowagi u człowieka. Zbiór ten obejmuje zagadnienia dotyczące zarówno teoretycznych aspektów kontroli postawy, w tym opracowanej nowej metodyki analizy danych stabilograficznych, jak i praktycznych zastosowań klinicznych.

Opracowane zostały metody badawcze wykorzystujące VR w analizach przygotowania posturalnego w reakcji na bodźce mogące potencjalnie doprowadzić do zaburzenia równowagi ciała. Zastosowano również wielkości bazujące na zmianach położenia COP, które nie mają charakteru cyklicznego do oceny zmian strategii utrzymywania równowagi. W badaniach stopniowo przechodzono od prostszych eksperymentów oceniających zdolność utrzymywania równowagi ciała w wirtualnej rzeczywistości [A1], przez wprowadzenie bardziej złożonych bodźców destabilizujących w świecie rzeczywistym [A2], aż po rozwój nowych metod analizy danych, które mają na celu uzupełnienie tradycyjnych technik oceny równowagi [A3, A4, A5], kończąc na próbie praktycznego wykorzystania opracowanych metod w praktyce klinicznej na grupie osób z chorobą Parkinsona [A6]. W badaniach wykorzystujących VR i symulację upadku ze schodów [A1] potwierdzono, że sam wizualny bodziec destabilizujący może mieć różny wpływ na równowagę badanych osób. Zauważono natomiast, że analiza częstotliwościowa, szczególnie ruchów głowy, dostarczyła bardziej precyzyjnych informacji o reakcji badanych na zadany wizualny bodziec, co sugeruje, że dodatkowe analizy mogą pomóc w lepszym wykrywaniu subtelnych zmian. W związku z tym konieczne były dalsze badania z bardziej wyraźnymi bodźcami destabilizującymi, w tym rzeczywistymi oraz wykorzystaniem pomiarów elektromiograficznych (EMG) i kinematyki całego ciała w celu poszerzenia prowadzenia analiz. Jako odpowiedź na wyżej wymienione potrzeby w drugim etapie badań [A2] zastosowano rzeczywiste zaburzenia, w postaci przesunięcia podłoża, które aktywowało mechanizmy przygotowania posturalnego (APA i EPA). Wyniki pokazały, że wiedza o nadchodzącym bodźcu wpływała na reakcję posturalną, szczególnie w zakresie wzrostu napięcia mięśniowego kończyn dolnych. Zrozumienie, w jaki sposób układ nerwowy przygotowuje się i reaguje na zewnętrzne bodźce jest kluczowe w kontekście diagnostyki i rehabilitacji. W tym przypadku pomocne mogłoby okazać się okazać się zastosowanie analiz zwiększających możliwości interpretacyjne sygnału mierzonego podczas oceny zdolności utrzymywania równowagi. Z tego powodu w następny etap badań [A3] dotyczył rozszerzenia tradycyjnych metod analizy równowagi o nowe podejścia, inspirowane technikami z analizy technicznej trendów giełdowych. Wyznaczono współczynnik Trend Change Index (TCI). Współczynnik ten definiuje liczbę zmian trendu, określanych jako liczba przecięć sygnału wynikających z algorytmu obliczeniowego Moving Average Convergence Divergence (MACD). MACD jest przedstawiany w postaci dwóch linii: linii MACD oraz linii sygnałowej. Przecięcie linii MACD i linii sygnałowej sygnalizuje zmianę trendu w sygnale przemieszczenia COP. Współczynnik TCI jest zatem sumą liczby zmian trendu sygnału w trakcie pomiaru [A3 – A5]. W celu przetestowania i dalszego rozwoju metody analizy zmian trendu sygnału COP w kolejnym etapie badano reakcję osób w warunkach zaburzeń wygenerowanych w technologii wirtualnej rzeczywistości [A4] i podczas rzeczywistego zaburzenia w postaci przesunięcia podłoża [A5]. Analiza zmian trendu wykazała, że liczba zmian trendu, oraz czas i odległość pomiędzy kolejnymi punktami oznaczającymi zmianę trendu w sygnale nimi mogą wpływać na zdolność utrzymywania stabilnej postawy ciała i sugerować zmianę strategii utrzymywania równowagi. Wnioski płynące z badań pozwoliły na przypuszczenia, że analiza zmian trendu może być przydatna w diagnostyce i ocenie osób dotkniętych chorobami neurodegeneracyjnymi. Z tego powodu ostatnim krokiem badań była próba praktycznego wykorzystania analizy zmian trendu w czasie analizy zdolności utrzymywania równowagi u osób z chorobą Parkinsona (PD) [A6]. W kontekście chorób takich jak choroba Parkinsona, analiza trendów okazała się przydatna w wykrywaniu różnic między stanami "on" i "off" leczenia dopaminergicznego. Pozwala ona na detekcję zmian stabilności posturalnej, które nie są widoczne w tradycyjnych pomiarach, co sugeruje jej potencjalne zastosowanie w monitorowaniu postępów choroby oraz w diagnostyce zaburzeń neurologicznych. Wykorzystanie analizy trendów u pacjentów z PD pozwala też na bardziej dokładne monitorowanie postępów choroby, co może być kluczowe w dostosowywaniu leczenia i rehabilitacji.

2.3 Analiza przemieszczeń COP i ruchów głowy w odpowiedzi na bodziec generowany w wirtualnej rzeczywistości [A1]

Zaburzenia równowagi mogą być sygnałem oznaczającym choroby neurologicznych lub wynikiem procesu starzenia. Standardowe testy zdolności utrzymywania równowagi, takie jak pomiary przemieszczeń COP, często są niewystarczające do wykrycia subtelnych zaburzeń równowagi. Z tego względu rośnie potrzeba wprowadzenia nowych, bardziej zaawansowanych metod, które byłyby w stanie wykryć zmiany, które są niewidoczne w tradycyjnych analizach w dziedzinie czasu. W tym kontekście technologia wirtualnej rzeczywistości (VR) odgrywa coraz większą rolę, umożliwiając symulowanie różnorakich warunków destabilizujących. Dzięki technologii VR możliwe jest wywołanie destabilizującego bodźca wizualnego, przy jednoczesnym utrzymaniu nieruchomego podłoża, co wprowadza badaną osobę w sytuację konfliktu sensorycznego. W literaturze opisywane są badania zdolności utrzymywania równowagi w VR podczas ekspozycji na ruchome otoczenie czy też symulacji sytuacji dnia codziennego. Analiza reakcji na tak przedstawione zaburzenie wizualne może być źródłem informacji na temat mechanizmów utrzymywania stabilnej postawy ciała. W badaniach zdolności utrzymywania równowagi najczęściej analizowane są wielkości bazujące na wielkościach w dziedzinie czasu. Jednakże, mimo wzrostu wartości wielkości w dziedzinie czasie w warunkach konfliktu bodźców sensorycznych, nie zawsze są one bezpośrednio związane z zaburzeniami równowagi związanymi np. z chorobami układu nerwowego. Z tego powodu rośnie zapotrzebowanie na nowe metody pomiarów i analiz, które uwydatniałyby zmiany niezauważalne w standardowej analizie w dziedzinie czasu. Analiza częstotliwościowa i zastosowanie akcelerometrów uzupełniają standardową analizę o dodatkowe informację dotyczące np. cykliczności ruchów COP czy charakterystyki ruchów poszczególnych segmentów ciała.

Z tego powodu cele pracy badawczej w pierwszym artykule cyklu obejmowały:

- określenie czy bodziec wizualny, w postaci symulowanego upadku ze schodów ma wpływ na kontrolę posturalną,

- określenie czy rozszerzenie analiz o analizę częstotliwościową może zwiększyć zakres interpretacji zdolności utrzymywania równowagi w porównaniu do analiz opartych na domenie czasowej,
- określenie czy pomiary ruchów głowy mogą uzupełnić pomiary COP o informacje dotyczące wpływu wprowadzonego wizualnego bodźca wytrącającego z równowagi na zdolność utrzymania równowagi,
- określenie, czy wizualny sygnał ostrzegawczy poprzedzający wystąpienie wizualnego bodźca wytrącającego z równowagi wywoła zmianę w ruchach głowy i przemieszczeniu COP.

W celu odpowiedzi na powyższe pytania zaprojektowano serię badań, w której wzięło udział 10 uczestników (7 kobiet i 3 mężczyzn) o średniej wieku 25 lat i średnim BMI 23 kg/m^2. Uczestnicy nie deklarowali poważnych urazów kończyn dolnych ani problemów z równowagą. Trzy osoby z grupy badawczej (jedna osoba z silną chorobą lokomocyjną i lękiem wysokości (pp3), oraz dwie osoby (pp1, pp2) o zwiększonych parametrach, które mogły świadczyć o problemach z utrzymaniem równowagi) zdecydowano się wyłączyć z grupy osób zdrowych i przeprowadzić osobne analizy dla każdego przypadku. Badanie przeprowadzono przy użyciu platformy pomiarowej WinFDM-S oraz zestawu VR HTC Vive. Aplikacja VR w Unity 3D przedstawiała prostą scenerię w postaci pokoju, a awatar osoby badanej został umiejscowiony na podłodze przed schodami prowadzącymi w dół (Rysunek 1). Podczas 60 sekundowych pomiarów mierzono COP i ruchy głowy. Testy w VR były poprzedzone badaniem stania na platformie z otwartym (EO) i zamkniętymi oczami (EC) w świecie rzeczywistym. W pierwszym teście, w 30 sekundzie symulowano upadek ze schodów (BB). W drugim teście uczestnicy otrzymywali wizualny sygnał ostrzegawczy na 3 sekundy przed symulacją (BZ). Każdy test powtórzono trzykrotnie.

Rysunek 1 Sceneria VR [A1]

Analiza wyników składała się z trzech etapów. W pierwszym etapie porównane zostały wartości prędkości przemieszczeń COP (V_{COP}), ruchów głowy (V_{head}) oraz powierzchni elipsy COP (EA_{COP}) w środowisku rzeczywistym i VR, w celu zidentyfikowania wpływu destabilizującego bodźca wzrokowego. Następnym krokiem, była analiza zakresu ruchu COP i zakresu ruchów głowy (DAD) w kierunku AP. Ostatnim etapem analiz było studium przypadku porównujące parametry osób z trudnościami w utrzymaniu równowagi z osobami bez takich problemów. Analiza statystyczna została przeprowadzona w oprogramowaniu Statistica 13. Ze względu na brak rozkładu normalnego we wszystkich wielkościach do porównania różnic wykorzystano testy nieparametryczne.

Aby zbadać jak symulowany upadek ze schodów może wywołać utratę równowagi i czy analiza przemieszczeń COP oraz ruchów głowy może umożliwić identyfikację problemów związanych z równowagą analizy przeprowadzono w dwóch kierunkach - analiza zmian parametrów opisujących destabilizację i kompensację posturalną dla grupy zdrowych osób oraz porównanie wyników dla osób, u których stwierdzono zwiększony wpływ wprowadzonego wizualnego bodźca na równowagę, z wynikami osób zdrowych.

Analiza zmian parametrów opisujących destabilizację i kompensację posturalną dla grupy zdrowych osób wykazała, że wprowadzony bodziec destabilizujący nie miał znaczącego wpływu na zachowanie uczestników. Wyniki nie wykazały różnic w reakcjach uczestników z i bez sygnału ostrzegawczego. Prawdopodobnym powodem tej sytuacji był fakt, że bodziec destabilizujący w formie symulowanego upadku ze schodów był wyłącznie wizualny i jego efekt był prawdopodobnie na tyle nierealistyczny, że nawet informacja o jego pojawieniu się i wcześniejsze doświadczenie tego, jak to będzie wyglądać, nie zmieniły reakcji uczestników. Podczas pomiarów zidentyfikowano trzy osoby, jedną z deklarowaną silną chorobą lokomocyjną i lękiem wysokości i dwie, u których wartości wielkości opisujące zdolność utrzymywania równowagi znacznie różniły się od reszty. Te różnice mogą wskazywać na problemy z utrzymaniem równowagi w określonych warunkach, jak przebywanie na wysokości czy niespodziewany ruch podłoża. W badaniach nie stwierdzono istotnych różnic w średniej prędkości COP i głowy między zdrowymi osobami a trzema osobami podatnymi na zaburzenia, co sugeruje, że te parametry nie są wystarczające do identyfikacji problemów z równowagą. Wyraźniejsze różnice zauważono w polu elipsy COP i ruchów głowy, gdzie wartości dla trzech osób były wyższe niż dla reszty grupy, co może wskazywać na trudności w utrzymaniu równowagi po zaburzeniu. Analiza częstotliwościowa wykazała, że bodziec w postaci symulowanego upadku ze schodów wywołał u badanych osób reakcję. W zmierzonym sygnale przebiegu COP w kierunku AP wykryto składową cykliczną o największej amplitudzie wynoszącej od 0,1 Hz do 0,2 Hz. Analiza tej składowej dla grupy kontrolnej oraz osób pp1, pp2 i pp3 nie ujawniła różnic między testami z ostrzeżeniem i bez ostrzeżenia, jednak zauważono znaczące różnice między trzema osobami (pp1, pp2, pp3) a resztą uczestników. Największe różnice dotyczyły amplitudy pierwszej harmonicznej, szczególnie w odniesieniu do ruchów głowy, gdzie wartości te były znacznie wyższe niż u reszty grupy. Te różnice wskazują na przyjęcie strategii balansowania ciałem, gdzie ruchy głowy dominują nad przemieszczeniami COP, co może być istotne w diagnozowaniu problemów z równowagą.

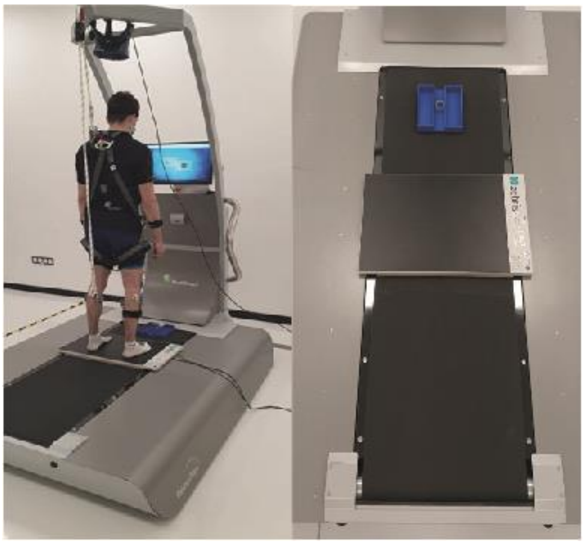
Podsumowując, bodźce wizualne generowane w wirtualnej rzeczywistości mają różny na osoby badane. Symulacja upadku ze schodów miała za zadanie wytworzenie sytuacji do zbadania reakcji ciała człowieka na nagłe, wizualne zaburzenie równowagi. W tym przypadku stworzona symulacja nie wpłynęła znacznie na kontrolę posturalną u badanych osób. Konieczne wydaje się przetestowanie bodźców wizualnych, które w większym stopniu wpływają na stabilność posturalną. Analiza w domenie częstotliwościowej, szczególnie analiza amplitudy pierwszej harmonicznej w sygnale przemieszczenia głowy, lepiej różnicowała osoby podatne na zewnętrzne zaburzenia wizualne. Chociaż takie wyniki nie zawsze wskazują na problemy zdrowotne, mogą sugerować zwiększoną podatność na nieoczekiwane zachowania w sytuacjach wymagających szybkich ruchów głowy. Aby w pełni zrozumieć mechanizmy odpowiedzialne za reakcje na zewnętrzne bodźce destabilizujące konieczna jest dodatkowa analiza z wykorzystaniem elektromiografii (EMG) oraz rozszerzenie analizy kinematyki ruchów ciała człowieka z samej głowy na całe ciało, a w szczególności na kończyny dolne oraz staw kolanowy. Wyniki badań podkreślają potrzebę dalszego dopracowywania metod symulowania wizualnych bodźców destabilizujących oraz metod analizy zdolności utrzymywania równowagi i reakcji na wizualne bodźce, która wykrywałaby zmiany dotąd niewykrywalne w standardowej analizie w dziedzinie czasu i częstotliwości.

2.4 Wpływ aktywności wybranych mięśni kończyny dolnej na poziom zaburzeń równowagi w reakcji na bodziec wytrącający z równowagi [A2]

Niniejszy artykuł stanowi kontynuację wcześniejszych badań nad analizą odpowiedzi motorycznej na zaburzenia wizualne, ale wprowadza nowe podejście poprzez zastosowanie rzeczywistych zaburzeń równowagi w postaci bodźca destabilizującego w postaci przesuwu podłoża. Mechanizmy Antycypacyjnego (APA) i Wczesnego (EPA) Przygotowania Posturalnego odgrywają kluczową rolę w dostosowaniu ciała do reakcji na bodźce zewnętrzne, umożliwiając utrzymanie równowagi i stabilności posturalnej. W ostatnich latach badania nad APA i EPA zyskały na znaczeniu, zwłaszcza w kontekście diagnozowania zaburzeń posturalnych oraz prognozowania upadków. Obserwacja tych mechanizmów pozwala na głębsze zrozumienie mechanizmów kontrolujących reakcje posturalne. Zastosowanie narzędzi takich jak platforma stabilograficzna, system do elektromiografii oraz czujniki IMU pozwala na bardziej kompleksowe i dokładne monitorowanie zarówno aktywności mięśniowej, przemieszczeń COP jak i kinematyki całego ciała. Niniejszy artykuł bada praktyczne zastosowania tych mechanizmów w ocenie stabilności posturalnej i ich potencjalne wykorzystanie w rehabilitacji oraz prewencji urazów.

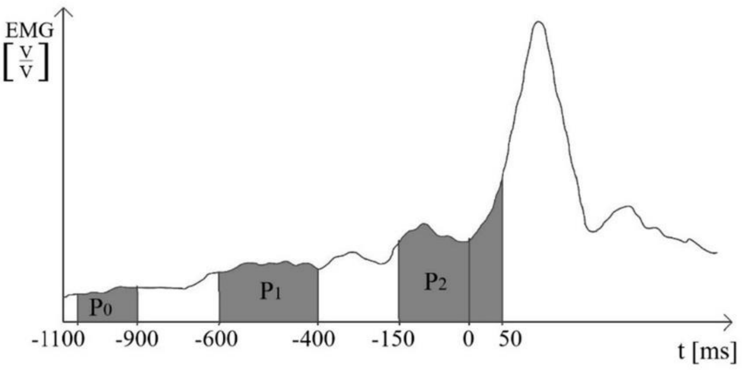
Celem tej pracy jest zbadanie czy informacja o czasie wystąpienia rzeczywistego bodźca zaburzającego równowagę wpływa na napięcie mięśni kończyn dolnych przed wystąpieniem zaburzenia. Do szczegółowych celów należało wykrycie występowania zjawiska APA i EPA, określenie, czy zwiększona aktywność mięśniowa jest ciągła w czasie, czy też nagła i krótko poprzedza zaburzenie oraz czy napięcie mięśni kończyn dolnych na początkowym etapie związanym z EPA prowadzi do zwiększenia napięcia mięśniowego w fazie związanej z APA. Dodatkowo, określono czy wzrost napięcia mięśni kończyn dolnych przed zaburzeniem wynika ze zmian posturalnych wywołujących przesunięcie środka masy do przodu lub do tyłu, zwiększenia lub zmniejszenia nacisku przodostopia na podłoże czy ze zwiększeniem kąta zgięcia w stawie kolanowym.

W celu odpowiedzi na powyższe cele zaprojektowano serię badań, w której wzięło udział 38 uczestników (27 kobiet i 11 mężczyzn) o średniej wieku 23 lat, średnim wzroście 172 cm i średniej masie 70 kg. Uczestnicy deklarowali brak historii poważnych urazów kończyn dolnych oraz dysfunkcji układu ruchu czy zaburzeń równowagi. Stanowisko pomiarowe składało się z platformy mierzącej nacisk stóp (WinFDM-S), bieżni do treningu i oceny posturalnych perturbacji (BalanceTutor), umożliwiającej wprowadzenie destabilizujących przemieszczeń podłoża w kierunku AP i ML, bezprzewodowego zestawu do elektromiografii (Ultium EMG) oraz czujników IMU (Ultium Motion). Platforma stabilograficzna była umieszczona centralnie na bieżni i sztywno przymocowana. Jeden czujnik IMU umieszczony został w specjalnie zaprojektowanym uchwycie i położony na pasie bieżni przed platformą (Rysunek 2). Wszystkie systemy były zsynchronizowane za pomocą oprogramowania Noraxon MR3 i platformy M5stack z mikrokontrolerem ESP32, co umożliwiało szybkie wykrycie ruchu bieżni.



Rysunek 2 Stanowisko pomiarowe [A2]

Do badań elektromiograficznych na podstawie przeglądu literatury zostały wybrane cztery mięśnie o kluczowych znaczeniu w kontekście utrzymywania równowagi: mięśniu piszczelowym przednim (TA), mięśniu prostym uda (RF), mięśniu brzuchatym łydki przyśrodkowym (GM) i bocznym (GL). Testy te są często stosowane do badania APA i CPA. Elektrody powierzchniowe zostały umieszczone na skórze w pobliżu brzuśców mięśni i połączono z bezprzewodowymi sensorami EMG. Dodatkowo, w celu identyfikacji kątów zgięcia w stawach, użyto 17 sensorów IMU, zawierających akcelerometr, żyroskop i magnetometr, rozmieszczonych na tułowiu, kończynach i głowie. Procedura testowa składała się z dwóch etapów: spoczynek (ERx) i zaburzenia (Tr). Podczas etapu spoczynku uczestnik siedział na krześle, trzymając stopy płasko na podłodze i rozluźniając mięśnie kończyn dolnych przez 15 sekund, w celu zarejestrowania ich aktywności. W drugim etapie uczestnik miał za zadanie stać nieruchomo, z twarzą skierowaną do przodu i rękami swobodnie opuszczonymi wzdłuż ciała na platformie stabilograficznej na bieżni, zabezpieczony specjalną uprzężą. Test obejmował trzy próby. Każda próba składała się z dwóch ruchów bieżni – do przodu i do tyłu. Pierwszy ruch był zawsze do przodu i był inicjowany 10 sekund po rozpoczęciu pomiarów. Ruch do tyłu był inicjowany 20 sekund po rozpoczęciu pomiarów. Zarówno ruchy do przodu, jak i do tyłu bieżni wynosiły 9,5 cm i trwały 0,52 sekundy. W pierwszej próbie (Tr1) uczestnik nie znał charakteru, czasu ani kierunku zaburzenia. W drugiej próbie (Tr2) znał charakter zaburzenia, ale nie czas ani kierunek. W trzeciej próbie (Tr3) uczestnik znał czas (odliczanie) i kierunek zaburzenia. Wyznaczono następujące wartości: t0 – czas początku ruchu oraz EMGRx – średnia aktywność mięśni podczas spoczynku. W celu standaryzacji aktywność każdego mięśnia mierzona w Tr1, Tr2 i Tr3 została podzielona przez wartość EMGRx. Następnym krokiem było zbadanie aktywności mięśni w określonych przedziałach czasowych odpowiadających kolejno obszar aktywacji mięśni podczas swobodnego stania (P0), obszar poszukiwania wzrostu aktywności mięśniowej wywołanej przez EPA (P1), obszar wzrostu aktywności mięśniowej wywołanej przez APAs (P2) (Rysunek 3).



$$P_0 = \int_{-1100}^{-900} EMG\,dt,$$

$$P_1 = \int_{-600}^{-400} EMG\,dt,$$

$$P_2 = \int_{-150}^{50} EMG\,dt,$$

$$EMG_{EPA} = P_1 - P_0,$$

$$EMG_{EPA} = P_2 - P_0.$$

Rysunek 3 Analizowane przedziały czasowe, równania do wyznaczania aktywności mięśniowej [A2]

Wartości EMG_{APA} i EMG_{EPA} zostały zidentyfikowane dla każdego z mięśni poddanych testom. Następnie skorelowano wyżej wymienione wartości w odniesieniu do wszystkich analizowanych mięśni w teście Tr3, obliczono wartość nacisku przodostopia na podłoże oraz wartości kąta zgięcia w stawie kolanowym w odniesieniu do testu Tr3. Następnie sprawdzono czy zwiększona aktywność mięśniowa związana z APA wpływa na przemieszczenie i prędkość COP po utracie równowagi. Analiza statystyczna została przeprowadzona Matlab R2022a przy pomocy testu ANOVA Friedmana oraz test post-hoc Wilcoxona z korektą Holma. Ze względu na brak normalnych rozkładów w zbiorach uzyskanych danych, obliczono korelację Spearmana.

Uzyskane wyniki pozwoliły na zauważenie wzrostu napięcia mięśniowego tylko wtedy, gdy badana osoba znała moment wystąpienia zaburzenia. W badaniach próbowano zdefiniować strategię przygotowania posturalnego oraz jej wpływ na kompensację posturalną po zaburzeniu. Różnice istotne statystycznie w EMG_{APA} i EMG_{EPA} między testami Tr1 i Tr3 oraz testami Tr2 i Tr3 zaobserwowano tylko w odniesieniu do ruchu do przodu dla TA, GL i GM w analizach podczas występowania zjawiska APA oraz dla TA i GM (tylko lewa kończyna dolna) w analizach dla zjawiska EPA. Statystycznie istotne korelacje między EPA a APA wskazywały, że mięśnie TA i GM były aktywowane wcześniej i ich aktywność wzrastała aż do momentu zaburzenia. Nie stwierdzono statystycznie istotnych różnic w średnich wartościach nacisku przodostopia ani kąta zgięcia w kolanie między testami Tr1, Tr2 i Tr3. Zaobserwowano korelację między zwiększoną aktywnością mięśniową a prędkością oraz maksymalnym przemieszczeniem COP po zaburzeniu, szczególnie dla mięśnia TA w ruchu do przodu bieżni oraz dla mięśnia GL w ruchu do tyłu. W obu przypadkach obserwowany wzrost napięcia mięśniowego skutkował wydłużeniem ścieżki i prędkości przemieszczenia COP.

Badania miały na celu zidentyfikowanie wpływu przesunięcia podłoża do przodu i do tyłu na reakcję mięśni kończyn dolnych związaną z dostosowaniem posturalnym poprzez ocenę aktywności wybranych mięśni, ocenę zgięcia w stawie kolanowym oraz ocenę nacisku wywieranego na podłoże, co było równoznaczne z przemieszczeniem COP. Występowanie zjawiska APA i EPA zostało zidentyfikowana na podstawie aktywności mięśniowej. Analiza wybranych mięśni wykazała, że podczas fazy APA w odniesieniu do Tr3, tj., gdy uczestnicy testu byli poinformowani o czasie rozpoczęcia zaburzenia, można było zauważyć wzrost średniej aktywności mięśniowej. Wykazano, że aktywność mięśnia TA odgrywała kluczową funkcję w adaptacji związanej z zaburzeniem posturalnym. Ze względu na brak różnic istotnych statystycznie pomiędzy próbami Tr1, Tr2, Tr3 w wartości zgięcia w stawie kolanowym, przemieszczenia COP oraz nacisku przodostopia można przypuszczać, że dostosowanie posturalne do zaburzenia nie było związane z pochyleniem tułowia do przodu ani do tyłu, a obserwowany wzrost napięcia TA, GL i GM nie wpływał na zgięcie w stawie kolanowym. Ze względu na obecność silnej korelacji pomiędzy napięciem mięśniowym a przemieszczeniem COP oraz prędkością COP po zaburzeniu oraz między napięciem mięśniowym TA a przemieszczeniem COP oraz prędkością przemieszczenia COP może wskazywać, że zwiększone napięcie mięśniowe w fazie APA było odpowiedzialne za blokadę stawów przed zaburzeniem, co z kolei zmieniało wzór kompensacji posturalnej po zaburzeniu, zwiększając zarówno długość ścieżki, jak i prędkość COP.

Badania jednoznacznie potwierdziły, że czynnikiem wywołującym dostosowanie posturalne i gotowość do reakcji na zaburzenia była wiedza o oczekiwanym czasie bodźca zaburzającego. Ważnym czynnikiem związanym z przygotowaniem do zaburzenia było napięcie mięśniowe odpowiedzialne za usztywnienie stawów kończyn dolnych, a w konsekwencji większy zakres ruchu COP po zaburzeniu. Połączenie rzeczywistych zaburzeń, takich jak niespodziewany i spodziewany ruch podłoża, z analizą EMG i IMU, daje możliwość precyzyjnej oceny mechanizmów utrzymywania równowagi. Takie podejście może dostarczyć nowych informacji na temat interakcji pomiędzy układem nerwowym a mięśniowym w procesie utrzymywania równowagi, co jest szczególnie istotne w kontekście prewencji upadków oraz rehabilitacji. Wyniki wskazują na konieczność szerszej analizy reakcji posturalnej na rzeczywiste bodźce destabilizujące, takie jak przesunięcie podłoża. Pomocne mogłoby okazać się wprowadzenie nowej metodyki analizy, który byłaby w stanie wykrywać niezauważalne w standardowej analizie zmiany niecykliczne. Zrozumienie, jak organizm przygotowuje się i reaguje na zewnętrzne bodźce wytrącające z równowagi, jest kluczowe dla opracowania lepszych metod diagnostyki i rehabilitacji, które mogłyby skuteczniej poprawiać stabilność posturalną i zmniejszać ryzyko upadków w codziennych sytuacjach.

2.5 Wskaźniki giełdowe w badaniach nad równowagą człowieka [A3]

Dotychczasowe badania dotyczące zdolności utrzymania równowagi najczęściej opierają się na pomiarach przemieszczeń środka nacisku stóp, takich jak prędkość COP, pole elipsy czy zakres przemieszczeń w różnych kierunkach. W przypadkach, gdy bodźce zaburzające równowagę mają charakter cykliczny, analizy w dziedzinie częstotliwości mogą dostarczyć cennych informacji na temat reakcji organizmu. Jednak analiza częstotliwościowa jest ograniczona do wykrywania cyklicznych zmian w pozycjach COP, pomijając zmiany niecykliczne, które również mogą być istotne dla oceny strategii utrzymania równowagi. W tym kontekście, nowe metody analizy, inspirowane technikami stosowanymi w analizie technicznej cen akcji na giełdzie, mogą rozszerzyć tradycyjne podejścia poprzez uwzględnienie chwilowych, niecyklicznych zmian COP i COM. Tego rodzaju zaawansowane podejście może nie tylko poprawić dokładność diagnostyczną, ale także umożliwić wcześniejsze wykrywanie zaburzeń równowagi, co jest szczególnie ważne w kontekście przeciwdziałania upadkom i związanym z nimi urazom. W trzecim artykule w cyklu opisana została metoda analizy zdolności utrzymywania równowagi oparta na wskaźniku giełdowym, która ma potencjał, aby uzupełnić zarówno analizę w dziedzinie czasu oraz częstotliwości podczas badań w wirtualnej rzeczywistości [A1] oraz podczas analizy przygotowania i kompensacji posturalnej [A2].

Artykuł przedstawia próbę wykorzystania wskaźnika Moving Average Convergence/Divergence (MACD), związanego z analizą techniczną trendów cen akcji w analizach dotyczących oceny zdolności utrzymania równowagi w świecie rzeczywistym i wirtualnej rzeczywistości.

Celem badania było wykazanie możliwości wykorzystania wskaźników giełdowych do oceny zdolności do utrzymania równowagi jako uzupełnienia analiz z wykorzystaniem wartości w dziedzinie czasu i częstotliwości. Wykorzystanie wskaźników giełdowych ma na celu umożliwienie wykrywania znaczących zmian w trendzie – zmian kierunku – w ruchu COP wraz z określeniem czasu między kolejnymi skokami. Taka analiza powinna umożliwić wykrywanie zarówno cyklicznych komponentów, jak i zmian niecyklicznych w określonym zakresie częstotliwości.

W badaniach wzięło udział 83 zdrowych uczestników (56 kobiet i 27 mężczyzn) o średniej wieku 21 lat, średnim wzroście 172 cm i średniej masie 65 kg. Uczestnicy deklarowali brak historii poważnych urazów kończyn dolnych oraz dysfunkcji układu ruchu czy zaburzeń równowagi. Badanie przeprowadzono przy użyciu platformy pomiarowej WinFDM-S oraz zestawu Oculus Rift. Aplikacja VR w Unity 3D przedstawiała dwie scenerie – „otwartą” pustynną scenę z obiektami widzianymi w odległości około 100 m oraz „zamkniętą” w postaci umeblowanego pokoju z obiektami w odległości około 3 m. Podczas pomiarów scenerie oscylowały w płaszczyźnie strzałkowej przy stałych częstotliwościach. Procedura badawcza obejmowała pomiary w środowisku rzeczywistym (z otwartymi i zamkniętymi oczami) oraz w środowisku wirtualnym z użyciem dwóch scenerii ("otwartej" i "zamkniętej") oscylujących z różnymi częstotliwościami (0,2 Hz, 0,5 Hz, 0,7 Hz i 1,4 Hz). Uczestnicy stali boso na platformie pomiarowej, z rękami skrzyżowanymi na piersiach i głową skierowaną na wprost. Pomiary obejmowały przemieszczenia COP podczas 30-sekundowych testów. Analizę wyników przeprowadzono w oprogramowaniu MATLAB. W pierwszym etapie analizowano średnią prędkość COP oraz zakres ruchu COP w kierunku AP. Następnie przeprowadzono analizę częstotliwościową, określając gęstość widmową (PSD) przemieszczenia COP w kierunku AP. Obliczono również współczynnik Trend Change Index (TCI), który definiuje liczbę zmian trendu na podstawie wskaźnika MACD.

Wskaźnik MACD jest przedstawiony w formie dwóch linii: linii MACD i linii sygnałowej. Linia MACD została uzyskana poprzez odjęcie szybkiej średniej kroczącej (średnia 26 okresowa) od wolnej średniej kroczącej (średnia 12 okresowa) (Równanie 1, Równanie 2).

Równanie 1

$$MACD_{12,26} = EMA_{12} - EMA_{26}$$

gdzie:

EMA_{12} – szybka średnia krocząca,

EMA_{26} – wolna średnia krocząca.

Równanie 2

$$EMA_{pN} = \frac{p_0 + (1-\alpha)p_1 + (1-\alpha)^2 p_2 + (1-\alpha)^3 p_3 + \cdots + (1-\alpha)^N p_N}{1 + (1-\alpha) + (1-\alpha)^2 + (1-\alpha)^3 + \cdots + (1-\alpha)^N}$$

gdzie:

p0 – bieżąca wartość sygnału,

pN – wartość sygnału poprzedzająca N okresów pomiarowych,

N – liczba okresów pomiarowych uwzględnianych w obliczeniach.

Linia sygnału została uzyskana poprzez obliczenie średniej wykładniczej o okresie 9 z linii MACD (Równanie 3).

Równanie 3

$$Signal\ Line = EMA_{MACD,9}$$

Przecięcie linii MACD i linii sygnału sygnalizuje zmianę trendu w sygnale przemieszczenia COP w kierunku AP. Zmiany trendu są związane ze zmianą kierunku ruchu COP. Współczynnik TCI określa liczbę zmian trendu w sygnale podczas 30-sekundowego testu, obliczaną jako suma przecięć linii MACD i linii sygnału.

Obliczone wartości współczynników PSD i TCI zostały użyte do analizy wpływu zaburzeń, na jakie byli narażeni uczestnicy. Porównano zarówno efekty scenerii i częstotliwości oscylacji na uczestników, jak i różnice w odniesieniu do środowiska rzeczywistego. Obliczenia zarówno dla PSD, jak i TCI zostały wykonane dla kierunku AP. Szczegółowa analiza wykazała, że odstępy czasowe między poszczególnymi zmianami trendu różniły się. W związku z tym wykryte zmiany trendu zostały pogrupowane według czasu poprzedzającego te zmiany. Przedziały czasowe zostały przekształcone na odpowiadające im częstotliwości, aby umożliwić porównawczą analizę współczynników MACD i PSD (Tabela 1).

Tabela 1 Przedziały czasowe brane pod uwagę podczas analiz oraz odpowiadające im przedziały częstotliwości [A3]

T [s]	0.05–0.1	0.1–0.2	0.2 – 0.3	0.3 – 0.4	0.4 – 0.5	0.5 – 0.6
f [Hz]	5.0-10.0	2,5-5.0	1,67-2,5	1,25-1,67	1.0-1,25	0,83-1.0
T [s]	0.6 – 0.7	0.7 – 0.8	0.8 – 0.9	0.9 – 1.0	0.05-1.0	1.0-30.0
f [Hz]	0,71-1.0	0,625-1	0,56-0,625	0,5-0,56	>0.5	<0.5

Analiza statystyczna wyników została przeprowadzona przy użyciu oprogramowania Statistica 13. Przeprowadzono test ANOVA Kruskala-Wallisa oraz test post-hoc Dunna, aby zbadać, czy występują istotne statystycznie różnice w analizowanych grupach.

Zmierzone wartości kolejnych pozycji COP w czasie zostały opracowane przy użyciu analiz w dziedzinie czasu i częstotliwości oraz na podstawie analizy trendu. W analizie w dziedzinie czasu (średnia prędkość COP, średnia prędkość COP w kierunku AP, zakres ruchu w kierunku AP) wykazano statystycznie istotny wzrost wartości wielkości mierzonych w środowisku wirtualnym w porównaniu do rzeczywistego dla większości pomiarów, z wyjątkiem średniej prędkości COP i zakresu ruchu w kierunku AP przy częstotliwościach oscylacji 0,2 Hz i 0,5 Hz. Obliczono współczynniki PSD i TCI dla kierunku AP. Wyniki analiz PSD wskazały na statystycznie istotne różnice przy porównaniu testów z otwartymi oczami (EO) do wszystkich innych pomiarów oraz w zakresie 0,5–10 Hz dla testów z zamkniętymi oczami (EC) w porównaniu do pomiarów w sceneriach "otwartej" i "zamkniętej" przy częstotliwościach 0,5 Hz, 0,7 Hz i 1,4 Hz. Nie stwierdzono istotnych różnic przy porównaniach pomiarów w wirtualnym środowisku między sobą zarówno w przypadku PSD jak i TCI. Porównania wartości PSD i TCI pokazały, że maksymalne wartości uzyskano przy częstotliwościach oscylacji scenerii 0,7 Hz i 1,4 Hz, niezależnie od typu scenerii. Statystycznie istotne różnice stwierdzono także przy porównaniu pomiarów w wirtualnym środowisku przy częstotliwościach 0,2 Hz i 0,5 Hz z pomiarami przy częstotliwości 1,4 Hz.

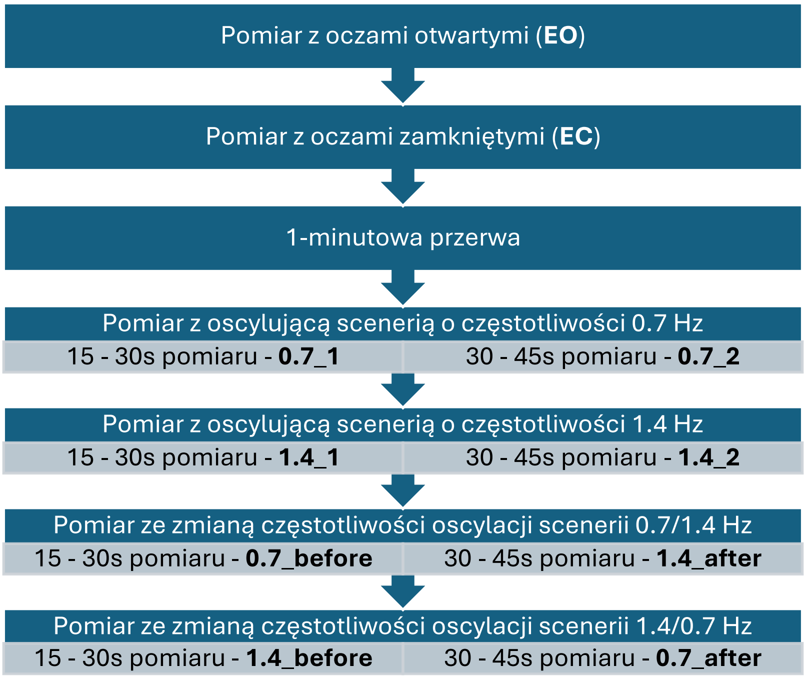
Tradycyjne wielkości używane do opisywania zdolności utrzymywania równowagi, takie jak długość ścieżki COP, średnia prędkość i zakres ruchu, mogą wzrosnąć u zdrowych osób przy niestabilnym podłożu lub w sytuacji konfliktu sensorycznego. W takich przypadkach analiza częstotliwościowa umożliwia rozkład sygnału ruchu COP na składniki cykliczne i określenie częstotliwości głównych składników ruchu. Jednakże, tradycyjne metody nie wykrywają niecyklicznych korekt pozycji COP. Proponowana w pracy analiza wykorzystująca współczynniki giełdowe umożliwiła wykrycie zmian trendu w ruchu COP, zarówno cyklicznych, jak i niecyklicznych. Zaproponowano obliczenia nowego współczynnika, który definiuje liczbę zmian trendu w sygnale przemieszczenia COP. Analiza liczby zmian trendu dla poszczególnych pomiarów pokazuje, że liczba tych zmian nie zależała od przetestowanych warunków badania. Może to oznaczać, że kontrola równowagi wymaga pewnej liczby szybkich zmian trendu ruchu COP (dla przedziału czasowego 0,05–1,0 s, co odpowiada częstotliwościom od 10 Hz do 0,5 Hz). Pogorszenie warunków, takie jak wprowadzenie zaburzenia w postaci oscylującej scenerii, nie zmienia znacząco tej liczby. Na tej podstawie można wnioskować, że albo układ równowagi nie potrzebuje większej liczby tego typu ruchów, albo układ ruchowy nie jest w stanie ich wykonać.

Przedstawione wyniki badań pokazują, że tradycyjne metody analizy, takie jak pomiar długości ścieżki COP, średniej prędkości czy zakresu ruchu, są skuteczne w ocenie ogólnej zdolności utrzymywania równowagi. Jednak mogą one nie w pełni wykrywać subtelne, niecykliczne zmiany w przemieszczeniu COP. Proponowana metoda analizy z wykorzystaniem analizy trendu opartej na wskaźniku giełdowym pozwala na wykrycie szybkich, niecyklicznych korekt postawy, co może dostarczyć nowych informacji diagnostycznych, szczególnie istotnych w kontekście badania osób z zaburzeniami równowagi. Niezbędny jest dalszy rozwój tej metody oraz jej testowanie w sytuacji występowania różnorodnych bodźców, zarówno wizualnych w środowisku wirtualnej rzeczywistości, jak i rzeczywistych, symulujących codzienne życie. Szczególnie ważne wydaje się przetestowanie tej metody na grupie osób dotkniętych np. chorobami neurodegeneracyjnymi, gdzie subtelne zmiany w równowadze mogą być kluczowe dla wczesnej diagnostyki i monitorowania postępu choroby. Integracja tej metody w praktyce klinicznej może prowadzić do bardziej precyzyjnej oceny zdolności utrzymania równowagi oraz lepszego zrozumienia mechanizmów kontroli proprioceptywnej.

2.6 Wpływ zaburzeń wizualnych na zmiany trendów w przebiegach przemieszczeń COP [A4]

Czwarty artykuł w cyklu jest bezpośrednią kontynuacją badań wprowadzających analizę trendów w badaniach zdolności utrzymywania równowagi [A3] i jednocześnie odpowiedzią na potrzebę dalszego rozwoju tej metody w celu poszerzenia możliwości prowadzenia analiz wyników pomiarów prowadzonych w środowisku VR z zaburzeniami. Artykuł miał na celu sprawdzenie czy analiza z wykorzystaniem analizy zmian trendu może skutecznie uzupełniać standardowe metody oceny równowagi poprzez identyfikację liczby korekt posturalnych (TCI) oraz wprowadzenie dodatkowych wskaźników opartych na TCI. Ponadto, badanie miało na celu określenie czy zmniejszenie częstotliwości korekt posturalnych oraz zwiększenie odległości między punktami zmiany trendu mogą być wskaźnikami zwiększonego ryzyka upadku, co mogłoby znaleźć zastosowanie w diagnostyce i ocenie równowagi.

W badaniach wzięło udział 28 zdrowych uczestników (13 kobiet i 15 mężczyzn) o średniej wieku 22 lat, średnim wzroście 173 cm i średniej masie 68 kg. Kryteriami wykluczającymi z badań były problemy zdrowotne związane z utrzymaniem równowagi i błędnikiem oraz otyłość (wskaźnik masy ciała BMI > 30). Badanie przeprowadzono przy użyciu platformy pomiarowej WinFDM-S oraz zestawu Oculus Rift. Scenerie trójwymiarowe zostały opracowane w środowisku Unity 3D. Sceneria „zamknięta” przedstawiała umeblowany pokój, w którym obiekty były widziane przez badanego w odległości około 3 m, natomiast sceneria „otwarta” przedstawiała pustynię, gdzie obiekty znajdowały się w odległości około 100 m. Podczas testów sceneria oscylowała w kierunku AP z stałą częstotliwością. Procedura badawcza obejmowała testy w rzeczywistym środowisku, polegające na staniu z otwartymi oczami (EO) i zamkniętymi oczami (EC), a także testy w środowisku wirtualnym.



Rysunek 4 Procedura badawcza [A4]

Pomiary w VR były przeprowadzane z użyciem scenerii otwartej i zamkniętej oscylującej z częstotliwościami 0,7 Hz, 1,4 Hz oraz ze zmianą częstotliwości w połowie badania z 0,7 Hz na 1,4 Hz i 1,4 Hz na 0,7 Hz (Rysunek 4). Uczestnicy stali boso na platformie pomiarowej, z rękami skrzyżowanymi na piersiach i głową skierowaną na wprost. Każdy pomiar trwał 60 sekund. W analizach uwzględniono okres od 15 do 45 sekundy, czyli czas, w którym dla badań z oscylującą scenerią trwały oscylacje.

Dane pomiarowe zostały przetworzone w oprogramowaniu MATLAB. Analiza obejmowała przemieszczenia COP w kierunku AP w czasie pomiarów EO i EC, oraz w trakcie środkowych 30 sekund testów z oscylacjami scenerii przy częstotliwościach 0,7 Hz i 1,4 Hz oraz ze zmianą częstotliwości (15 sekund przed i 15 sekund po zmianie). Obliczono podstawowe wartości stabilograficzne, takie jak prędkość COP i zakres ruchu COP w kierunku AP oraz wskaźnik zmiany trendu (TCI). Wskaźnik TCI został przedstawiony jako sumaryczna liczba wszystkich zmian trendu dla analizowanego pomiaru oraz łączna liczba zmian trendu podzielona na następujące okresy czasowe: 0–0,2 s, 0,2–0,5 s oraz 0,5–1 s. Każdy z okresów czasowych określa czas, który upłynął po wystąpieniu jednej zmiany trendu do wystąpienia kolejnej. Na podstawie algorytmu TCI obliczono również następujące wartości: średnią odległość między kolejnymi punktami zmiany trendu (MACD_dS), średni czas między kolejnymi punktami zmiany trendu (MACD_dT) oraz średnią prędkość zmian przemieszczeń między kolejnymi punktami zmiany trendu (MACD_dV).

Do analizy statystycznej wyników, ze względu na brak rozkładu normalnego, użyto testów ANOVA Friedmana oraz post hoc Wilcoxona z korekcją Holma.

W czasie analizy wyniki podzielono na trzy grupy: wartości standardowe w dziedzinie czasu, wartości wskaźnika zmiany trendu (TCI) w całości pomiaru oraz z podziałem na okresy czasowe oraz wartości obliczone na podstawie analizy trendu (MACD_dT, MACD_dS, MACD_dV). Wartości prędkości COP w kierunku AP wzrosły statystycznie istotnie po zamknięciu oczu przez uczestnika oraz po wprowadzeniu zburzeń wizualnych w VR. W przypadku wartości TCI porównanie pomiarów EO i EC nie wykazało istotnych statystycznie różnic. Istotne statystycznie różnice zaobserwowano przy porównywaniu testu EO w środowisku rzeczywistym z pomiarami w środowisku wirtualnym w przypadku oscylującej scenerii o częstotliwości 0.7 Hz. Wprowadzenie zaburzeń do VR zmniejszyło wartości mediany TCI w przedziale czasowym 0-0.2s. Dla przedziału czasowego 0.2–0.5 s można zauważyć, że wyższe wartości mediany występowały w pomiarach z użyciem scenerii oscylującej o częstotliwości 1.4 Hz. Dla przedziału czasowego 0.5–1 s najwyższe wartości zaobserwowano w testach VR z zaburzeniami o częstotliwości 0.7 Hz. Analizując wartości MACD_dS można zauważyć znaczny wzrost uzyskanych wartości w testach VR w porównaniu z pomiarami wykonanymi w środowisku rzeczywistym. Przy porównaniu pomiarów uzyskanych w VR między poszczególnymi testami nie stwierdzono istotnych statystycznie różnic. W przypadku analizy wartości MACD_dT znacznie wyższe wartości uzyskano dla pomiarów przeprowadzonych przy częstotliwości 0.7 Hz. Wartości mediany MACD_dV wzrosły po zamknięciu oczu oraz po wprowadzaniu wirtualnej rzeczywistości.

Otrzymane wyniki wykazują, że w większości przypadków wartość TCI była zbliżona, niezależnie od warunków pomiarowych, co może sugerować, że do utrzymania równowagi konieczna jest pewna liczba korekt postawy. Analizowane wartości MACD_dS, MACD_dT i MACD_dV dostarczyły dodatkowych informacji o ruchu COP. Wartości te mogą wskazywać, czy zmiany prędkości wynikają ze zmian długości skoków COP, czasu ich trwania, czy obu tych czynników jednocześnie. W badaniach zaobserwowano, że wartość MACD_dT, która nie ulega znaczącemu spadkowi, jest istotna z perspektywy zdolności utrzymania równowagi, a jednoczesny wzrost MACD_dS i spadek MACD_dT świadczący o dłuższych przeskokach COP odbywających się w krótszym czasie może prowadzić do destabilizacji i ewentualnego upadku badanej osoby.

Wykorzystanie wskaźników giełdowych do oceny stabilności ciała ludzkiego uzupełnia standardowe analizy w dziedzinie czasu i częstotliwości. Analiza wartości TCI, MACD_dV, MACD_dT i MACD_dS dostarcza dodatkowych informacji o czynnikach wpływających na standardowe wielkości opisujące zdolność utrzymywania równowagi, takie jak długość ścieżki, średnia prędkość i zakres ruchu. Łącząc analizę zmian trendu z analizą wielkości stabilograficznych, możemy uzyskać informacje na temat częstotliwości korekcji postawy, odstępów między korekcjami i prędkości ruchu COP.

2.7 Analiza zmian trendu w ocenie równowagi ciała podczas korekt postawy w reakcji na zaburzenie [A5]

Piąty artykuł w cyklu stanowi połączenie badań nad reakcjami posturalnymi w odpowiedzi na zewnętrzy bodziec wytrącający z równowagi [A2] z innowacyjną metodą analizy zmian trendu do zdolności utrzymywania równowagi [A3, A4]. Tradycyjne metody analizy postawy, takie jak pomiary aktywności mięśni i przemieszczeń środka nacisku stóp, mogą być uzupełnione przez analizy zmian trendu (TCA). TCA, inspirowana metodami stosowanymi w analizie giełdowej, umożliwia identyfikację szybkich korekt posturalnych i analizę niecyklicznych zmian w sygnale COP, co może dostarczyć nowych informacji na temat strategii utrzymania równowagi i reakcji na destabilizujące bodźce. Artykuł jednocześnie jest odpowiedzią na potrzebę dalszego rozwoju i testowania analizy zmian trendu, w tym przypadku, w środowisku rzeczywistym, przy występowaniu rzeczywistego bodźca wytrącającego z równowagi. W niniejszym artykule została postawiona hipoteza, że wzrost prędkości COP może być wynikiem zmiany strategii utrzymania równowagi, co powinno znaleźć odzwierciedlenie w zmianach parametrów analizy trendu w sygnale COP, takich jak liczba zmian trendu oraz czas i odległość pomiędzy nimi. Celem badania było ustalenie, czy różne warunki podczas badania zjawisk związanych z przygotowaniem posturalnym wpływają na wartości wielkości uzyskanych w czasie analizy trendu.

W badaniu wzięło udział 38 osób (27 kobiet, 11 mężczyzn), o średnie wieku 23 lata, średnim wzroście 172 cm oraz średniej masie ciała 70 kg. Czynnikami wykluczającymi z badań były przebyte urazy kończyn dolnych oraz problemy z równowagą. Stanowisko pomiarowe składało się z platformy mierzącej nacisk stóp (WinFDM-S), bieżni do treningu i oceny posturalnych perturbacji (BalanceTutor), umożliwiającej wprowadzenie destabilizujących przemieszczeń podłoża w kierunku AP i ML, bezprzewodowego zestawu do elektromiografii (Ultium EMG) oraz czujników IMU (Ultium Motion). Platforma stabilograficzna była umieszczona centralnie na bieżni i sztywno przymocowana. Elektrody oraz czujniki systemu do elektromiografii zostały przymocowane na powierzchni skóry we okolicach brzuśców: mięśnia piszczelowego przedniego (TA), mięśnia prostego uda (RF), mięśnia brzuchatego łydki (głowa przyśrodkowa) oraz mięśnia brzuchatego łydki (głowa boczna) (GM) - mięśni aktywnie zaangażowanych w proces utrzymywania równowagi w kierunku przednio-tylnym. Jeden czujnik IMU został przymocowany do pasa bieżni, co umożliwiało wykrycie rozpoczęcia perturbacji w postaci poruszenia platformy na pasie bieżni i synchronizację wszystkich urządzeń (Rysunek). W pierwszej próbie uczestnicy nie wiedzieli, kiedy i w jakim kierunku nastąpi perturbacja (Tr1). W drugiej wiedzieli, że perturbacja nastąpi, ale nie znali jej dokładnego czasu ani kierunku (Tr2). W trzeciej próbie byli dokładnie informowani o czasie i kierunku perturbacji, z odliczaniem wyświetlanym na ekranie (Tr3). Podczas każdej próby platforma wykonywała dwa ruchy – do przodu i do tyłu – każdy o długości 9,5 cm i trwający 0,5 sekundy.

Rysunek 5 Stanowisko pomiarowe wraz z osobą badaną [A5]

Po wykonaniu pomiarów analizowano wartości w dziedzinie czasu, a także przeprowadzano analizę zmian trendu. W czasie analizy obliczono średnią prędkości COP oraz wielkości opisujące charakterystykę zmian trendu – całkowitą liczbę zmian trendu podczas całego testu (TCI), średnią odległość między kolejnymi punktami zmiany trendu (TCI_dS), średni czas między kolejnymi punktami zmiany trendu

(TCI_dT) oraz średnią prędkość chwilowych przemieszczeń pomiędzy kolejnymi zmianami trendu (TCI_dV). Test Shapiro-Wilka wykazał brak normalnych rozkładów dla wszystkich wielkości, dlatego w analizie statystycznej zastosowano mediany oraz testy nieparametryczne. Wielkości otrzymane w testach Tr1, Tr2 i Tr3 porównano za pomocą testu Friedmana oraz testu par Wilcoxona z poprawką Holma.

Analiza sygnałów EMG wykazała różnice w aktywacji mięśni kończyn dolnych między próbami, szczególnie w próbie Tr3, gdzie zaobserwowano wcześniejsze przygotowanie do zaburzenia równowagi (EPA i APA) [A2]. Badania potwierdziły, że różna wartość aktywacji mięśni wpływa na charakterystykę sygnałów COP, co było widoczne w zmianach wartości parametrów takich jak TCI_dS i TCI_dV. W szczególności, w próbie Tr3, na sekundę przed zaburzeniem odnotowano znaczący wzrost odległości między kolejnymi punktami oznaczającymi zmianę trendu, co sugeruje zmianę strategii utrzymania równowagi. Wzrost ten był również widoczny w liczbie przemieszczeń COP w określonych przedziałach czasowych. Wykazano, że w sytuacjach, gdy badani wiedzieli o nadchodzącym zaburzeniu, wzrastała prędkość COP (V_{COP}), co mogło wskazywać na zmianę strategii utrzymania równowagi. Mimo wzrostu V_{COP}, liczba zmian trendu (TCI) pozostała stała, co sugeruje, że zmiany te wynikały z wydłużenia odległości pokonywanych przez COP, a nie z częstszych korekt postawy. Zauważono również, że zwiększone napięcie mięśniowe, wynikające ze świadomości nadchodzącego ruchu, prowadziło do usztywnienia ciała, co mogło powodować zmiany w charakterystyce utrzymania równowagi. Badania wskazują, że reakcje ciała na zaburzenia równowagi są złożone i mogą się różnić w zależności od poziomu świadomości badanych na temat nadchodzącego zaburzenia.

Metoda analizy zmian trendu może uzupełniać tradycyjne analizy COP oraz pomiary EMG w badaniach nad przygotowaniem posturalnym (PA). Umożliwia ona określenie wielkości bazujących na TCI, które pomagają wyjaśniać przyczyny np. wzrostu prędkości COP, co ilustruje mechanizmy strategii utrzymania równowagi.

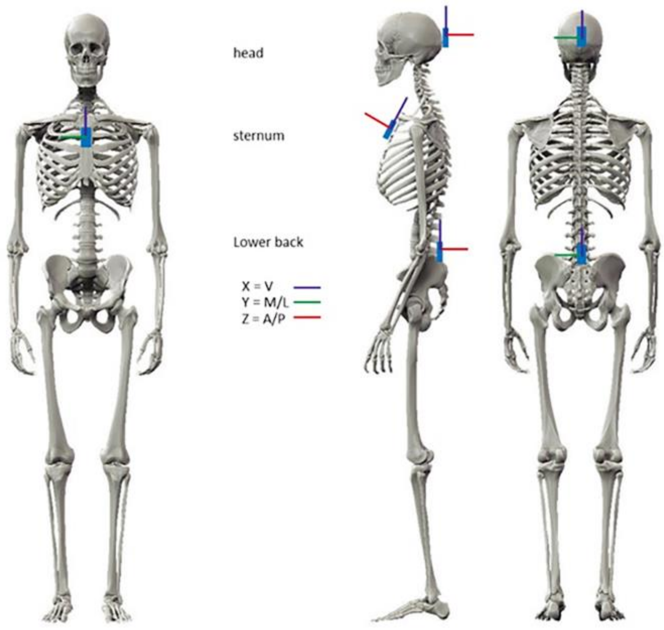
2.8 Analiza zmian trendu w równowadze posturalnej w chorobie Parkinsona [A6]

Szósty artykuł cyklu jest opisem wykorzystania innowacyjnej metody analizy trendu w badaniach zdolności utrzymywania równowagi z wykorzystaniem IMU w grupie osób z chorobą Parkinsona (PD). Badania te mogą przyczynić się w przyszłości do powszechnego stosowania analizy trendu w praktyce klinicznej, umożliwiając bardziej precyzyjne monitorowanie postępów choroby oraz tworzenie nowych metod diagnostyki w oparciu o szczegółowe analizy stabilności posturalnej pacjentów. Artykuł powstał we współpracy z Uniwersytetem w Kilonii, gdzie zostały wykonane pomiary.

Badania przedstawione w artykule mają dwa cele - zbadanie potencjalnego zastosowania analizy zmian trendu (TCA) do oceny stabilności posturalnej z wykorzystaniem IMU oraz zastosowanie tej analizy w kontekście chorób neurologicznych, w szczególności PD. Hipotezą badań jest, że TCA może różnicować osoby z chorobą Parkinsona (pwPD) i zdrowe osoby dorosłe, a także rozróżniać pomiędzy fazami "on" (PDon) i "off" (PDoff) związanymi z przyjmowaniem leków dopaminergicznych.

Grupa badawcza składała się z 61 zdrowych osób podzielonych na dwie podgrupy: młodych dorosłych (40 osób) (YO) o średniej wieku 30 lat, średnim wzroście 185 cm i średniej masie ciała 80 kg i starszych dorosłych (21 osób) (OP) o średniej wieku 73 lat, średnim wzroście 181 cm i średniej masie ciała 84 kg oraz 29 osób z chorobą Parkinsona (pwPD). Wśród podgrupy z chorobą Parkinsona 13 osób uczestniczyło w badaniu jako PDoff (ocena UPDRS III wynosiła 24 ± 10), 23 jako PDon (ocena UPDRS III wynosiła 30 ± 20), a 7 zarówno jako PDon (ocena UPDRS III wynosiła 26 ± 10), jak i PDoff (ocena UPDRS III wynosiła 27 ± 10). Wszyscy uczestnicy badań z chorobą Parkinsona byli pacjentami oddziału neurogeriatrycznego w Centrum Neurologii Uniwersyteckiego Szpitala Schleswig-Holstein w Kiel.

Trzy czujniki IMU (Ultium Motion) zostały przymocowane do ciała (na wysokości miednicy, mostka i głowy) każdej badanej osoby za pomocą elastycznych pasków (Rysunek). Uczestników poproszono o stanie w pozycji wyprostowanej z nogami razem, jedna obok drugiej i skupienie wzroku na punkcie na białej ścianie przez 10 s, jako część testu Short Physical Performance Battery.



Rysunek 6 Umiejscowienie czujników IMU [A6]

Dane z czujników IMU zostały przetworzone w oprogramowaniu MATLAB. Wyznaczono następujące wielkości dla ruchów COM: zryw (JERK) (cm^2/s^5), pole elipsy podparcia (SURFACE) (cm^2), ścieżka (PATH) (cm), średnia prędkość (MV) (cm/s), zakres przyspieszenia (RANGE) (cm/s^2) oraz błąd średniokwadratowy (RMS) (cm/s^2). Wykonana została również analiza zmian trendu [A3, A4] dla sygnału przyspieszenia, w wyniku której otrzymano wartości TCI, TCI_dT, TCI_dS oraz TCI_dV. Analiza statystyczna została przeprowadzona przy użyciu oprogramowania Matlab R2022a oraz JASP. Test Shapiro-Wilka wykazał brak normalnych rozkładów dla wszystkich wielkości, dlatego w analizie statystycznej zastosowano testy nieparametryczne. Do porównania różnic pomiędzy grupami i pozycjami czujników wykorzystano test Kruskala-Wallisa oraz test post-hoc Dunna z korekcją Bonferroniego.

Wyniki potwierdziły, że wykrywanie różnic w ruchach między segmentami ciała jest możliwe zarówno przy użyciu wielkości opisujących zdolność utrzymywania równowagi, jak i TCA, jednakże tylko TCA umożliwiła rozróżnienie stanów "on" i "off" u osób z PD. Wyniki sugerują, że podczas prostych zadań równoważnych osoby z PD mogą prezentować podobne zachowanie jak osoby zdrowe, co może być wynikiem kompensacyjnych mechanizmów w centralnym układzie nerwowym. Badanie wykazało, że leczenie dopaminergiczne może wpływać na zmiany w stabilności postawy. Wyniki TCA wskazały na większą liczbę zmian trendu (TCI) i mniejsze wartości TCI_dT w stanie "off" w porównaniu do stanu "on", co może odzwierciedlać deficyty wizualne wynikające z niedoboru dopaminy.

Najważniejszym wnioskiem płynącym z badań jest to, że wprowadzenie analizy zmiany trendu (TCA) okazało się kluczowe w wykrywaniu znaczących różnic między stanem "on" a "off" leczenia PD, co podkreśla jej potencjał w ocenie zmian związanych z chorobą, które nie są wykrywane za pomocą konwencjonalnych parametrów.

2.9 Podsumowanie

W ramach rozprawy doktorskiej przedstawiono wyniki badań opublikowanych w szeregu artykułów naukowych, które koncentrują się na mechanizmach kontroli posturalnej i zdolności utrzymywania równowagi ciała człowieka w różnych warunkach eksperymentalnych, w tym w rzeczywistym i wirtualnym środowisku.

Uzyskane wyniki badań dostarczyły informacji, że standardowo stosowane wielkości liczone w dziedzinie czasu, opisujące zdolność utrzymywania równowagi są niewystarczające do określenia reakcji badanej osoby na zadane bodźce. Zastosowanie analiz w dziedzinie częstotliwości może znacznie poszerzyć możliwości interpretacji otrzymywanych danych pomiarowych, co wykazano w jednym z badań podczas oceny wpływu ruchu głowy na utrzymywanie równowagi ciała. Zastosowanie tych analiz umożliwiło wyodrębnienie z grupy badawczej osób najbardziej podatnych na zaburzenia wizualne. W badaniach dotyczących wpływu rzeczywistych bodźców wytrącających z równowagi na występowanie zjawisk przygotowania posturalnego potwierdzono, że kluczowym czynnikiem, który wywoływał zjawiska przygotowania posturalnego i gotowość do reakcji na zaburzenie była informacja o czasie nadchodzącego zaburzenia. Dodatkowo wykazano, że połączenie analizy stabilograficznej, EMG i IMU poszerza możliwość precyzyjnej oceny mechanizmów utrzymywania równowagi. Wyniki badań potwierdziły, że metoda analizy chwilowych korekt postawy wzbogaca standardowe analizy w dziedzinie czasu i częstotliwości umożliwiając uwzględnienie niecyklicznych przemieszczeń COP. Wykazano przydatność tej metody zarówno podczas badań w środowisku wirtualnym z oscylującymi zaburzeniami jak i w środowisku rzeczywistym przy realnych bodźcach destabilizujących. Rozwój analiz w oparciu o detekcję chwilowych korekt postawy otwiera nowe możliwości w diagnostyce zaburzeń równowagi, szczególnie w odniesieniu do chorób neurodegeneracyjnych, jak choroba Parkinsona. Zastosowanie tych narzędzi w praktyce klinicznej pozwoliło na różnicowanie pomiędzy stanem "on" a "off" leczenia dopaminergicznego w PD. Wyniki te wskazują na duży potencjał zastosowanej metody, w tym do wczesnej diagnostyki zmian w stabilności posturalnej, które nie są widoczne przy użyciu tradycyjnych metod oceny równowagi.

Podsumowując, przeprowadzone badania umożliwiły realizację zakładanych celów badawczych. Zastosowanie badań z bodźcami destabilizującymi, zarówno rzeczywistymi jak i wygenerowanymi w technologii wirtualnej rzeczywistości znacząco poszerza możliwości analizy zdolności utrzymywania równowagi. Wykorzystanie metod detekcji chwilowych korekt postawy, zwiększa możliwości interpretacyjne zjawisk towarzyszących destabilizacji ciała osoby badanej. Tego typu podejście umożliwia bardziej precyzyjne monitorowanie stabilności posturalnej, co ma kluczowe znaczenie zarówno w rehabilitacji, jak i w prewencji upadków, a także w diagnostyce chorób neurodegeneracyjnych.

3. Poszerzone streszczenie w języku angielskim

3.1 Introduction

The ability to maintain balance is a key element of daily human functioning. It affects independence in performing everyday activities, mobility, and overall quality of life. Balance maintenance is a complex process involving the vestibular, visual, and proprioceptive systems. The vestibular system, located in the inner ear, comprises the semicircular canals, utricle, and saccule, which respond to head acceleration [1]. The visual system provides information about the surrounding environment, including light intensity, object positions within sight, and the location of visible body parts. The proprioceptive system supplies sensory data about the relative positions of body parts. Proprioceptor sensory endings respond to deformation, and several groups of proprioceptors play a crucial role in movement control. They are sensitive to physical variables such as joint position, muscle length, contraction speed, and muscle force. The integration of information from these systems is essential for maintaining balance. Disorders in any of these systems can lead to balance control difficulties and an increased risk of falls. Impaired balance may result from neurological diseases, aging, vestibular disorders, sensory disturbances in the feet due to diabetes, or musculoskeletal injuries [1,2]. Disrupted balance can lead to falls, which, by causing injuries, may further worsen health conditions [3]. Falls and related injuries are among the leading causes of hospitalization in older adults, often resulting in loss of mobility and independence [4]. Therefore, monitoring and improving balance, especially in the elderly and those with chronic illnesses, is crucial for enhancing quality of life and reducing fall risks.

Maintaining a stable posture is a dynamic process that requires constant control and adjustments of the center of mass to sustain balance during everyday activities. The process of balance maintenance involves mechanisms related to postural preparation (PA) and postural compensation in response to destabilizing stimuli. Postural preparation occurs just before a disturbance and aims to prevent or minimize the negative effects of balance loss. In contrast, compensatory adjustments are designed to restore balance immediately after the disturbance [5, 6, 7]. Postural control mechanisms can be categorized as early (early postural adjustment – EPA), which occur approximately 600–400 ms before the disturbance, preparing the body for the anticipated imbalance; anticipatory (anticipatory postural adjustment – APA), occurring approximately 150 ms before to 50 ms after the disturbance, preparing to counter the imbalance; or compensatory (compensatory postural adjustment – CPA), occurring approximately 70–300 ms after the disturbance, restoring balance [8, 9, 10]. These mechanisms primarily manifest as changes in the activity of postural muscles [8], the displacement of individual body segments, and shifts in the center of mass (COM) or center of pressure (COP) [6]. Stimuli triggering PA can be divided into two types: the first related to voluntary movement initiation [11], and the second external, originating from the environment, often leading to destabilization [5]. Research suggests that an external stimulus can trigger APA if the subject is aware of when the disturbance will occur [8, 10]. In rehabilitation and therapy, understanding these mechanisms is critical for identifying specific deficits in a patient's balance control system [12, 13], reducing the risk of falls [4, 14], and monitoring rehabilitation progress [15].

Scientists use various methods to assess the ability to maintain balance, with the most employed approach involving the analysis of center of pressure (COP) displacements. Metrics describing the ability to maintain a stable posture, based on COP displacements, are most frequently analyzed in the time domain [16, 17] and the frequency domain [18, 19]. The most popular time-domain metrics include COP path length, average COP velocity, COP ellipse area, and COP movement ranges in the anterior-posterior (AP) and medio-lateral (ML) directions. Assessments may also involve determining the position of the center of mass (COM), tracking head movements, and measuring trunk acceleration values [20]. Less frequently used methods, as described in the literature, include wavelet transform analysis [21, 22, 23] and probabilistic analyses based on entropy [24].

Frequency domain analyses complement time-domain analyses by identifying cyclic components in the COP signal. These analyses can reveal subtle changes that may be difficult to detect using standard time-domain metrics. Frequency domain approaches are particularly valuable in developing new diagnostic methods, especially for assessing balance abilities under sensory conflict conditions, both in real-world environments and through virtual reality (VR) applications. VR technology has gained popularity in recent years due to technological advancements, making it more accessible. The use of 3D imagery enables the creation of environments and stimuli that are challenging to replicate in the real world. Additionally, the wide range of customizable scenarios allows the diagnostic and rehabilitation process to be tailored to individual patient cases and abilities. As a result, VR is increasingly being applied in rehabilitation and diagnostics, particularly for conditions related to impaired balance. Research in sensory conflict scenarios, where different senses receive conflicting information, is especially relevant in this context. In such cases, participants are exposed to stimuli that directly or indirectly destabilize their posture. For example, a study might involve a person standing on stable ground while VR technology creates the visual illusion of a moving environment (e.g., by displaying oscillating surroundings) [18, 25, 26, 27]. Analyzing the body's response to these visual disturbances can offer valuable diagnostic insights into balance control and adaptability in dynamically changing environments. Impaired adaptability may serve as an early indicator of neurological disorders, such as Parkinson's disease or vestibular dysfunction [28].

Currently used methods for assessing balance abilities, such as time-domain [16, 17] and frequency-domain analyses of COP displacements [18, 19], offer valuable insights into postural stability. Commonly analyzed time-domain parameters include COP velocity, COP ellipse area, and COP displacement ranges in the anterior-posterior (AP) and medio-lateral (ML) directions. Typically, an increase in these parameters is interpreted as indicative of balance control issues [33]. In contrast, frequency-domain analyses facilitate the examination of cyclic components within the signal, identifying dominant frequencies, which aids in assessing changes in balance strategies. This approach is especially useful when studying the effects of cyclic disturbances in virtual reality environments [21, 22, 23], broadening the scope of balance assessments beyond what time-domain analyses provide. The incorporation of advanced technologies, such as IMU sensors, electromyography, and virtual reality, can further enhance the evaluation of balance abilities and postural preparation [29, 30, 31, 32]. However, these technologies require analyses that allow for a comprehensive understanding and interpretation of the data within the measured signal. While the methods described above detect changes in the signal that suggest alterations in balance strategies, they focus on global signal analysis or the cyclic nature of observed phenomena, often overlooking rapid, non-cyclic changes. These non-cyclic changes can be critical for diagnosing and treating patients with neurological, orthopedic, or vestibular disorders. Consequently, there is a need for the continued development of research methods that complement current measurements and analyses, providing a more complete understanding and interpretation of changes in balance abilities that may result from the progression of various diseases.

3.2 Research objective

The recognized need for developing measurement methods and exploring novel approaches to data analysis, aimed at improving our understanding of human balance control mechanisms, has led to the formulation of the following research objectives:

- Develop a methodology for measuring human balance abilities that assesses changes in postural control strategies in response to both anticipated and unanticipated destabilizing stimuli.
- Determine the impact of virtual and real destabilizing stimuli on postural preparation as a diagnostic tool for evaluating human balance abilities.
- Analyze the practical application of methods for detecting momentary postural corrections to assess changes in postural control strategies in response to virtual and real destabilizing stimuli.

This doctoral dissertation summarizes the results of a series of studies published in scientific articles, focusing on the mechanisms of postural control and the ability to maintain balance in humans. The work encompasses both theoretical aspects of postural control, including the development of new methodologies for analyzing stabilographic data, and practical clinical applications.

Research methods utilizing virtual reality (VR) were developed to analyze postural preparation in response to balance-disturbing stimuli. Additionally, non-cyclical changes in the center of pressure (COP) were measured to assess shifts in balance strategies. The studies evolved from simpler experiments evaluating balance in virtual reality environments [A1], to more complex real-world destabilizing stimuli [A2], and to the development of novel data analysis methods aimed at complementing traditional balance assessment techniques [A3, A4, A5]. The final phase involved applying these methods in clinical practice with patients diagnosed with Parkinson's disease (PD) [A6]. In studies involving VR and a simulated fall down the stairs [A1], it was confirmed that visual destabilizing stimuli affected participants' balance differently. Frequency analysis, particularly of head movements, provided more precise information about participants' responses to visual disturbances, suggesting that additional analyses could help detect subtle changes. This led to further studies using more pronounced destabilizing stimuli, including real-world perturbations and the integration of electromyography (EMG) and full-body kinematic measurements, to expand the scope of analysis. The second phase of research [A2] involved real-world disturbances, such as shifting ground, to activate postural preparation mechanisms (anticipatory postural adjustments—APA, and early postural adjustments—EPA). The results showed that awareness of an impending stimulus influenced postural reactions, particularly in terms of increased lower limb muscle tension. Understanding how the nervous system prepares for and reacts to external stimuli is crucial for diagnosis and rehabilitation. This phase highlighted the necessity for more advanced data analysis techniques to enhance balance assessment interpretation. The next research phase [A3] focused on extending traditional balance assessment methods by applying techniques inspired by stock market trend analysis. The Trend Change Index (TCI) was introduced, which defines the number of trend changes based on signal intersections from the Moving Average Convergence Divergence (MACD) algorithm. MACD is represented by two lines: the MACD line and the signal line. Their intersections indicate trend changes in COP displacement, and the TCI coefficient counts the number of these changes during measurement [A3 – A5]. To further validate the COP trend change analysis method, subsequent studies examined participants' responses to virtual reality-generated disturbances [A4] and real-world perturbations with shifting ground [A5]. The analysis showed that the number of trend changes, as well as the time and distance between them, influenced postural stability and indicated shifts in balance strategies. These findings suggest that trend change analysis may be valuable for diagnosing and assessing individuals with neurodegenerative diseases. The final phase of the research involved evaluating the practical application of trend change analysis in assessing balance in individuals with Parkinson's disease (PD) [A6]. In PD, trend analysis proved effective in detecting differences between "on" and "off" states during dopaminergic treatment, revealing changes in postural stability that were not detectable with traditional methods. This suggests that trend analysis may be valuable for monitoring disease progression and diagnosing neurological disorders. Additionally, trend analysis in PD patients allows for more precise tracking of disease progression, which could be critical for optimizing treatment and rehabilitation strategies.

3.3 Analysis of center of pressure displacements and head movements triggered by a visual stimulus created using the virtual reality technology [A1]

Balance disorders can indicate neurological diseases or result from the aging process. Standard balance tests, such as COP displacement measurements, are often insufficient for detecting subtle balance disturbances. As a result, there is an increasing need for more advanced methods capable of identifying changes that remain undetected by traditional time-domain analyses. In this context, virtual reality (VR) technology is playing an increasingly significant role, enabling the simulation of various destabilizing conditions. Through VR, it is possible to introduce destabilizing visual stimuli while maintaining stable ground, placing the individual in a situation of sensory conflict. Studies have described balance assessments in VR environments, often exposing participants to moving surroundings or simulations of everyday situations. Analyzing reactions to these visual disturbances provides valuable insights into the mechanisms underlying postural stability. In balance assessment studies, time-domain parameters are typically analyzed. However, while time-domain values often increase under sensory conflict conditions, they do not always correlate directly with balance disorders, particularly those caused by neurological conditions. This underscores the need for new measurement and analysis methods capable of detecting changes that are not visible in standard time-domain analysis. Frequency-domain analysis and the use of accelerometers complement traditional analyses by providing additional information, such as the cyclicity of COP movements or the characteristics of individual body segment movements.

Consequently, the research objectives of the first article in this series were as follows:

- determine whether a visual stimulus, in the form of a simulated fall down the stairs, affects postural control,
- assess whether incorporating frequency-domain analysis can enhance the interpretation of balance ability compared to time-domain analysis,

- evaluate whether head movement measurements can complement COP measurements by providing additional insights into the effects of visual destabilizing stimuli on balance ability,
- investigate whether a visual warning signal preceding the destabilizing stimulus influences head movements and COP displacement.

To answer the questions, a series of studies was designed involving 10 participants (7 women and 3 men) with an average age of 25 years and an average BMI of 23 kg/m^2. All participants reported no significant lower limb injuries or balance issues. However, three individuals - one with severe motion sickness and a fear of heights (pp3), and two others (pp1, pp2) with elevated parameters indicating potential balance issues - were excluded from the healthy group, and separate analyses were conducted for each case. The study was conducted using the WinFDM-S measurement platform and the HTC Vive VR set. The VR application, developed in Unity 3D, presented a simple room scenario where the participant's avatar stood at the top of a staircase leading downward (Figure 1). Throughout the 60-second trials, both COP and head movements were recorded. Prior to the VR tests, participants underwent balance assessments on the platform in a real-world setting, with eyes open (EO) and eyes closed (EC). In the first test, a simulated fall down the stairs was triggered at the 30-second mark (BB). In the second test, participants received a visual warning signal 3 seconds before the simulation (BZ). Each test was repeated three times.

Figure 1 VR scenery [A1]

The analysis of the results was conducted in three stages. In the first stage, comparisons were made between the values of COP displacement velocity (V_{COP}), head movement velocity (V_{head}), and COP ellipse area (EA_{COP}) in both real and VR environments to assess the impact of destabilizing visual stimuli. The second stage involved analyzing the range of COP and head movements (DAD) in the anterior-posterior (AP) direction. The final stage consisted of a case study comparing the parameters of individuals with balance difficulties to those without such issues. Statistical analysis was performed using Statistica 13 software. Since none of the variables followed a normal distribution, non-parametric tests were used to compare differences.

To determine whether the simulated fall down the stairs could induce balance loss and whether COP displacements and head movements could identify balance-related issues, the analysis was conducted in two ways: first, by evaluating changes in parameters describing destabilization and postural compensation in a group of healthy individuals, and second, by comparing the results of those more susceptible to visual disturbances with the healthy group.

The analysis of parameters describing destabilization and postural compensation in the healthy group revealed that the destabilizing visual stimulus did not significantly affect participants' behavior. No significant differences were found between their reactions with and without the warning signal. It is likely that the visual stimulus (a simulated fall downstairs) was too unrealistic to elicit substantial changes, even when participants were aware of its occurrence and had experienced it before. Among the subjects, three individuals—one with severe motion sickness and a fear of heights, and two others with elevated balance parameters—stood out. Their differing results suggest potential balance difficulties under specific conditions, such as being at heights or experiencing unexpected ground movement. The study did not find significant differences in average COP or head movement velocities between healthy individuals and those more susceptible to disturbances, indicating that these parameters alone may be insufficient for detecting balance problems. More noticeable differences were observed in COP ellipse area and head movements, where the values for the three individuals were higher than the rest of the group, suggesting greater difficulty in maintaining balance after a disturbance. Frequency analysis revealed that the visual stimulus triggered a cyclic component with the highest amplitude between 0.1 Hz and 0.2 Hz in the COP AP signal. While no differences were found between tests with and without the warning signal, significant differences were observed between the three individuals and the rest of the participants, particularly in the first harmonic amplitude related to head movements. These results suggest that head movements dominated over COP displacements in their balance strategy, which could be key in diagnosing balance problems.

In summary, visual stimuli in VR affect individuals differently. Although the simulated fall down the stairs was designed to assess postural responses to a sudden visual disturbance, it did not significantly impact the postural control of most participants. This suggests that stronger or more realistic visual stimuli are needed to assess postural stability effectively. Frequency domain analysis, particularly the first harmonic amplitude of head movement signals, was better at distinguishing individuals susceptible to visual disturbances. While these findings do not necessarily indicate health issues, they may suggest increased susceptibility to unexpected stimuli requiring rapid head movements. To fully understand the mechanisms behind responses to external destabilizing stimuli, further analysis using electromyography (EMG) and expanded kinematic assessments of body movements - especially those of the lower limbs and knee joints - is necessary. The findings highlight the need for further refinement of methods simulating visual destabilizing stimuli and balance assessment techniques that detect changes not captured by standard time and frequency domain analyses.

3.4 The effect of selected lower limb muscle activities on a level of imbalance in reaction on anterior-posterior ground perturbation [A2]

This article continues previous research on motor responses to visual disturbances, introducing a novel approach by applying real balance disruptions in the form of destabilizing ground shifts. Mechanisms of Anticipatory Postural Adjustment (APA) and Early Postural Adjustment (EPA) play a crucial role in adapting the body to external stimuli, helping to maintain balance and postural stability. In recent years, research on APA and EPA has become increasingly important, especially for diagnosing postural disorders and predicting falls. Observing these mechanisms provides deeper insights into how the body controls postural responses. Tools such as stabilographic platforms, electromyography systems, and IMU sensors offer more comprehensive and precise monitoring of muscle activity, COP displacements, and overall body kinematics. This article explores the practical applications of these mechanisms in assessing postural stability and their potential use in rehabilitation and injury prevention.

The aim of this study is to investigate whether knowledge of the timing of an actual balance-disrupting stimulus affects lower limb muscle tension before the disturbance occurs. Specific objectives include detecting the occurrence of APA and EPA, determining whether increased muscle activity is continuous over time or short-lived before the disturbance, and whether early muscle tension during the EPA phase leads to increased muscle tension in the APA phase. Additionally, the study aims to assess whether the increase in lower limb muscle tension before the disturbance results from postural changes that shift the center of mass forward or backward, increase or decrease forefoot pressure on the ground, or increase knee flexion angles.

To address the objectives, a series of studies was conducted involving 38 participants (27 women and 11 men) with an average age of 23 years, an average height of 172 cm, and an average weight of 70 kg. All participants reported no history of serious lower limb injuries, motor system dysfunctions, or balance disorders. The measurement setup consisted of a foot pressure measurement platform (WinFDM-S), a treadmill for postural perturbation training and assessment (BalanceTutor), which allowed for destabilizing displacements of the ground in both the anterior-posterior (AP) and medio-lateral (ML) directions, a wireless electromyography system (Ultium EMG), and IMU sensors (Ultium Motion). The stabilographic platform was centrally positioned on the treadmill and securely fixed in place. An IMU sensor was placed in a specially designed holder on the treadmill belt in front of the platform (Figure 2). All systems were synchronized using Noraxon MR3 software and an M5stack platform with an ESP32 microcontroller, enabling quick detection of treadmill movement.

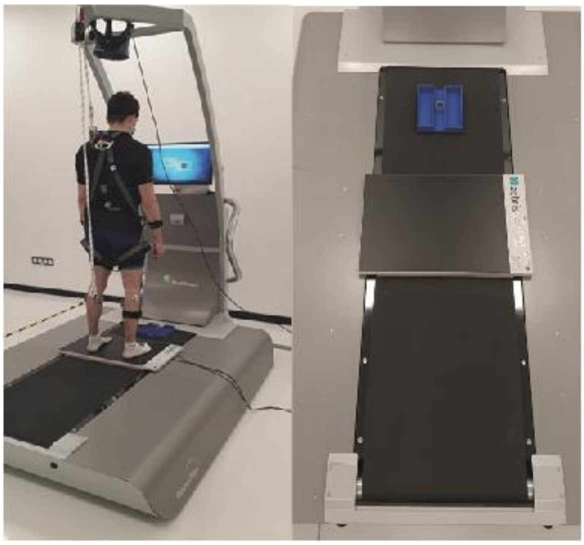
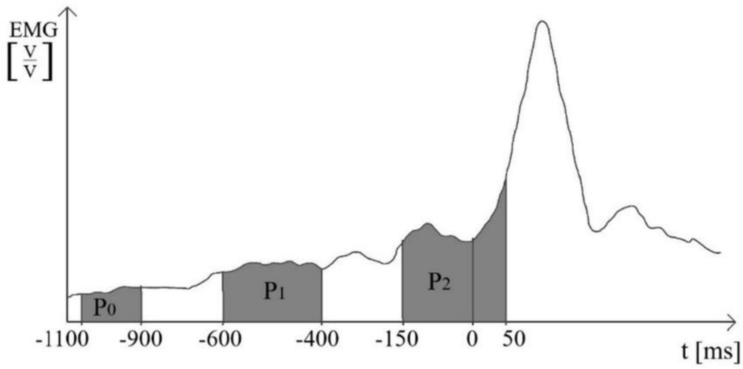


Figure 2 Measurement stand [A2]

Based on the literature review, four muscles were selected for the electromyographic studies due to their key roles in maintaining balance: the tibialis anterior (TA), rectus femoris (RF), medial gastrocnemius (GM), and lateral gastrocnemius (GL). These muscles are commonly examined in studies investigating Anticipatory Postural Adjustments (APA) and Compensatory Postural Adjustments (CPA). Surface electrodes were placed on the skin near the muscle bellies and connected to wireless EMG sensors. Additionally, 17 IMU sensors, each containing an accelerometer, gyroscope, and magnetometer, were used to track joint angles. These sensors were placed on the torso, limbs, and head. The test procedure consisted of two stages: rest (ERx) and perturbation (Tr). During the rest stage, participants sat on a chair with their feet flat on the floor, relaxing their lower limb muscles for 15 seconds to record baseline muscle activity. In the perturbation stage, participants stood still on the stabilographic platform on the treadmill, facing forward with their arms relaxed at their sides, secured by a safety harness. The test consisted of three trials, each involving two treadmill movements—forward and backward. The first movement was always forward, initiated 10 seconds after the start of the measurement, while the backward movement occurred 20 seconds after the start. Both forward and backward displacements covered 9.5 cm and lasted 0.52 seconds. In the first trial (Tr1), participants were unaware of the nature, timing, or direction of the perturbation. In the second trial (Tr2), they knew the nature of the perturbation but not its timing or direction. In the third trial (Tr3), participants were informed of both the timing (with a countdown) and the direction of the perturbation. The following values were determined: t0 – the start of the movement, and EMGRx – the average muscle activity during the rest phase. To standardize the data, muscle activity in Tr1, Tr2, and Tr3 was divided by the EMGRx value. Next, muscle activity was analyzed in specific time intervals corresponding to different phases of muscle activation: free standing (P0), the period anticipating increased muscle activity due to EPA (P1), and the period of increased muscle activity caused by APA (P2) (Figure 3).



$$P_0 = \int_{-1100}^{-900} \mathrm{EMG}\,dt,$$

$$P_1 = \int_{-600}^{-400} \mathrm{EMG}\,dt,$$

$$P_2 = \int_{-150}^{50} \mathrm{EMG}\,dt,$$

$$\mathrm{EMG}_{EPA} = P_1 - P_0,$$

$$\mathrm{EMG}_{EPA} = P_2 - P_0.$$

Figure 3 Time intervals and equations for determining muscle activity [A2]

EMGAPA and EMGEPA values were identified for each muscle assessed. These values were then correlated across all analyzed muscles during the Tr3 test. Additionally, forefoot pressure and knee flexion angle were calculated relative to the Tr3 test. The study then assessed whether increased muscle activity related to APA affected the displacement and velocity of the COP following balance loss. Statistical analysis was performed using Matlab R2022a, utilizing the Friedman ANOVA test and

Wilcoxon's post-hoc test with Holm's correction. Due to the non-normal distribution of the data, Spearman's correlation was also calculated.

The results showed an increase in muscle tension only when participants were aware of the timing of the disturbance. The study aimed to define the strategy of postural preparation and its impact on postural compensation after the disturbance. Statistically significant differences in EMGAPA and EMGEPA between Tr1 and Tr3, as well as between Tr2 and Tr3, were observed only for forward movement in the TA, GL, and GM muscles during APA, and for TA and GM (only in the left leg) during EPA. Significant correlations between EPA and APA indicated that the TA and GM muscles were activated earlier, with activity increasing until the disturbance occurred. No significant differences were found in forefoot pressure or knee flexion angle across the Tr1, Tr2, and Tr3 tests. A correlation between increased muscle activity and both COP velocity and maximum displacement after the disturbance was noted, particularly for the TA muscle during forward movement and GL during backward movement. In both cases, increased muscle tension led to extended COP path length and velocity.

The study aimed to evaluate the effect of forward and backward ground displacement on lower limb muscle responses related to postural adaptation by examining muscle activity, knee flexion, and pressure on the ground in relation to COP displacement. APA and EPA phenomena were identified based on muscle activity, and the analysis revealed an increase in average muscle activity during APA in Tr3, when participants were informed of the disturbance's timing. The TA muscle was shown to play a crucial role in adapting to postural disturbances. Due to the lack of significant differences in knee flexion, COP displacement, and forefoot pressure between Tr1, Tr2, and Tr3, it can be inferred that postural adaptation to the disturbance was not associated with forward or backward trunk leaning, and the observed increase in TA, GL, and GM muscle tension did not affect knee flexion. The strong correlation between muscle tension and COP displacement and velocity after the disturbance, especially between TA tension and COP behavior, suggests that increased muscle tension during the APA phase contributed to joint locking before the disturbance, thereby altering postural compensation and increasing both COP path length and velocity.

The study clearly confirmed that the key factor triggering postural adjustment and readiness for disturbance was awareness of the anticipated timing of the perturbation. Muscle tension, which stiffened the lower limb joints, was crucial for preparing for the disturbance, leading to greater COP displacement after the disturbance. Combining real-world disturbances, such as unexpected and anticipated ground shifts, with EMG and IMU analysis offers a precise assessment of balance control mechanisms. This approach provides new insights into the interaction between the nervous and muscular systems in maintaining balance, which is especially important for fall prevention and rehabilitation. The results emphasize the need for broader analysis of postural responses to real destabilizing stimuli, such as ground displacement. The introduction of a new analytical methodology capable of detecting non-cyclical changes, which standard analyses often miss, could be highly beneficial. Understanding how the body prepares for and responds to external balance disturbances is key to developing improved diagnostic and rehabilitation methods that can more effectively enhance postural stability and reduce the risk of falls in everyday situations.

3.5 The stock market indexes in research on human balance [A3]

Previous studies on balance maintenance ability have primarily focused on measuring center of pressure (COP) displacements, including parameters like COP velocity, ellipse area, and displacement range in different directions. When balance-disturbing stimuli are cyclical, frequency-domain analyses can provide valuable insights into the body's response. However, frequency analysis is limited to detecting cyclical changes in COP positions and may overlook non-cyclical changes that are also critical for assessing balance maintenance strategies. In this context, new analytical methods inspired by stock market price trend analysis techniques can expand traditional approaches by incorporating momentary, non-cyclical changes in COP and center of mass (COM) movements. These advanced approaches can not only enhance diagnostic accuracy but also allow for earlier detection of balance disturbances, which is particularly important for fall prevention and reducing injury risks. The third article in this series introduces a balance analysis method based on a stock market index, aiming to complement both time-domain and frequency-domain analyses in virtual reality studies [A1] and postural preparation and compensation analyses [A2].

This article explores the use of the Moving Average Convergence/Divergence (MACD) index - typically associated with stock price trend analysis - to evaluate balance maintenance ability in both real-world and virtual reality environments. The study aimed to demonstrate the feasibility of applying stock market indicators to assess balance ability, complementing traditional time and frequency-domain analyses. By detecting significant trend changes - directional shifts - in COP movement and identifying the time intervals between successive changes, this analysis could capture both cyclical and non-cyclical components within a specific frequency range.

The study involved 83 healthy participants (56 women and 27 men) with an average age of 21 years, an average height of 172 cm, and an average weight of 65 kg. All participants reported no history of significant lower limb injuries, motor system dysfunction, or balance disorders. The study was conducted using the WinFDM-S measurement platform and the Oculus Rift VR system. The VR application, developed in Unity 3D, presented two scenarios: an "open" desert scene with objects visible at approximately 100 meters, and a "closed" room scene with furniture and objects at about 3 meters. During the measurements, the scenes oscillated in the sagittal plane at fixed frequencies. The testing procedure included measurements taken in both real-world environments (with eyes open and closed) and in virtual environments using the two oscillating scenes ("open" and "closed") at different frequencies (0.2 Hz, 0.5 Hz, 0.7 Hz, and 1.4 Hz). Participants stood barefoot on the measurement platform with their arms crossed over their chest and heads facing forward. The measurements focused on COP displacements during 30-second tests. Data analysis was conducted using MATLAB software. In the first stage, the average COP velocity, and the range of COP movement in the AP direction were analyzed. This was followed by a frequency analysis to determine the Power Spectral Density (PSD) of COP displacement in the AP direction. Additionally, the Trend Change Index (TCI) was calculated to define the number of trend changes based on the Moving Average Convergence Divergence (MACD) index.

The MACD index is represented by two lines: the MACD line and the signal line. The MACD line was obtained by subtracting the slow-moving average (26-period average) from the fast-moving average (12-period average) (Equation 1, Equation 2).

Equation 4

$$MACD_{12,26} = EMA_{12} - EMA_{26}$$

where:

EMA_{12} – faster exponential moving average,

EMA_{26} – slower exponential moving average;

Equation 2

$$EMA_{pN} = \frac{p_0 + (1-\alpha)p_1 + (1-\alpha)^2 p_2 + (1-\alpha)^3 p_3 + \cdots + (1-\alpha)^N p_N}{1 + (1-\alpha) + (1-\alpha)^2 + (1-\alpha)^3 + \cdots + (1-\alpha)^N}$$

where:

p0 – ultimate value,

pN – value preceding N periods,

N – number of periods.

The signal line was obtained by calculating the moving average with exponential weight for the MACD signal considering the nine MACD signal samples (Equation 3).

Equation 3

$$Signal\ Line = EMA_{MACD,9}$$

The intersection of the MACD line and the signal line indicates a trend change in the COP displacement signal in the AP direction. These trend changes correspond to shifts in the direction of COP movement. The Trend Change Index (TCI) coefficient defines the number of trend changes in the signal during a 30-second test and is calculated as the total number of intersections between the MACD and signal lines.

The calculated PSD and TCI values were used to analyze the participants' responses to disturbances. The effects of different virtual scenarios and oscillation frequencies were compared, as well as the differences between the virtual and real-world environments. Calculations for both PSD and TCI were focused on the AP direction. A detailed analysis revealed that the time intervals between individual trend changes varied. As a result, these detected trend changes were grouped based on the time intervals preceding them. These time intervals were then converted into corresponding frequencies, enabling a comparative analysis between the MACD and PSD coefficients (Table 1).

Table 1 The time intervals considered during the analyzes [A3]

T [s]	0.05–0.1	0.1–0.2	0.2 – 0.3	0.3 – 0.4	0.4 – 0.5	0.5 – 0.6
f [Hz]	5.0-10.0	2,5-5.0	1,67-2,5	1,25-1,67	1.0-1,25	0,83-1.0
T [s]	0.6 – 0.7	0.7 – 0.8	0.8 – 0.9	0.9 – 1.0	0.05-1.0	1.0-30.0
f [Hz]	0,71-1.0	0,625-1	0,56-0,625	0,5-0,56	>0.5	<0.5

The statistical analysis of the results was conducted using Statistica 13 software. A Kruskal-Wallis ANOVA test and Dunn's post-hoc tests were performed to determine whether statistically significant differences existed among the analyzed groups.

The measured values of subsequent COP positions over time were processed using time, frequency domain analyses, and trend analysis. In the time-domain analysis (average COP velocity, average COP velocity in the AP direction, range of movement in the AP direction), a statistically significant increase in measured values in the virtual environment was observed compared to the real environment for most measurements. However, there were no significant differences for average COP velocity and range of movement in the AP direction at oscillation frequencies of 0.2 Hz and 0.5 Hz. PSD and TCI coefficients were calculated for the AP direction. The PSD analysis showed statistically significant differences when comparing tests with open eyes (EO) to all other measurements, as well as within the 0.5–10 Hz range for tests with closed eyes (EC) compared to the "open" and "closed" VR scenes at oscillation frequencies of 0.5 Hz, 0.7 Hz, and 1.4 Hz. No significant differences were found between the virtual environment tests themselves for both PSD and TCI values. Comparisons of PSD and TCI values indicated that the maximum values were obtained at oscillation frequencies of 0.7 Hz and 1.4 Hz, regardless of the scene type. Statistically significant differences were also observed when comparing the virtual environment tests at frequencies of 0.2 Hz and 0.5 Hz with those at 1.4 Hz.

Traditional metrics used to describe balance ability, such as COP path length, average velocity, and movement range, tend to increase in healthy individuals when standing on unstable surfaces or in situations of sensory conflict. In such cases, frequency analysis allows for the decomposition of the COP movement signal into cyclic components and identification of dominant movement frequencies. However, traditional methods often fail to detect non-cyclic corrections in COP position. The proposed analysis method using stock market indices enabled the detection of both cyclic and non-cyclic trend changes in COP movement. A new coefficient, the Trend Change Index (TCI), was introduced to define the number of trend changes in the COP displacement signal. The analysis of trend changes across different measurements showed that the number of these changes remained consistent across tested conditions. This suggests that balance control requires a certain number of rapid trend changes in COP movement (within time intervals of 0.05–1.0 s, corresponding to frequencies of 10 Hz to 0.5 Hz). Destabilizing conditions, such as the introduction of an oscillating visual scene, did not significantly alter this number. From this, it can be inferred that either the balance system does not require additional movements, or the motor system is unable to generate them.

The presented results show that traditional methods of analysis, such as measuring COP path length, average velocity, and movement range, are effective in assessing general balance ability but may not fully capture subtle, non-cyclic changes in COP displacement. The proposed trend analysis method, based on a stock market index, allows for the detection of rapid, non-cyclic postural corrections, which could provide new diagnostic insights, especially for individuals with balance disorders. Further development and testing of this method are necessary under a variety of stimuli, both in virtual reality environments and in real-world simulations that mimic everyday scenarios. It is especially important to assess this method on individuals with neurodegenerative diseases, where subtle changes in balance may be critical for early diagnosis and monitoring disease progression. Integrating this method into clinical practice could lead to more precise balance assessments and a better understanding of proprioceptive control mechanisms.

3.6 Impact of Visual Disturbances on the Trend Changes of COP Displacement Courses Using Stock Exchange Indices [A4]

The fourth article in the series builds on previous research introducing trend analysis in balance assessment [A3], while addressing the need for further development of this method to enhance the analysis of measurement results in a VR environment with disturbances. The article aimed to determine whether trend change analysis (TCI) can effectively complement standard balance assessment methods by identifying the number of postural corrections (TCI) and introducing additional indicators based on TCI. Furthermore, the study sought to determine whether a reduction in the frequency of postural corrections and an increase in the distance between trend change points could serve as indicators of increased fall risk, which could be applied in balance diagnostics and evaluation.

The study involved 28 healthy participants (13 women and 15 men) with an average age of 22 years, an average height of 173 cm, and an average weight of 68 kg. Exclusion criteria included health problems related to balance or the vestibular system, as well as obesity (body mass index BMI > 30). The experiment was conducted using the WinFDM-S measurement platform and the Oculus Rift VR system. The 3D scenes were developed in the Unity 3D environment. The "closed" scene depicted a furnished room with objects visible to the participant at approximately 3 meters, while the "open" scene portrayed a desert landscape with objects located about 100 meters away. During the tests, the scenes oscillated in the AP direction at a constant frequency. The experimental procedure included tests in a real-world environment, where participants stood with eyes open (EO) and eyes closed (EC), as well as tests conducted in the virtual environment.

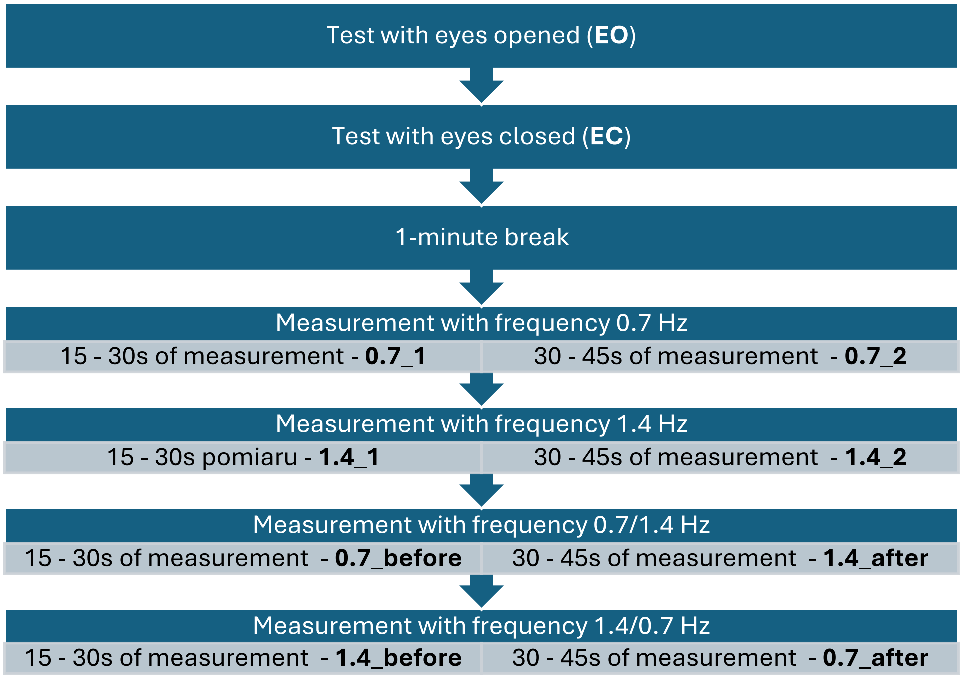


Figure 4 Experimental procedure [A4]

Measurements in VR were conducted using both open and closed scenes oscillating at frequencies of 0.7 Hz and 1.4 Hz, as well as with frequency changes midway through the test—from 0.7 Hz to 1.4 Hz and from 1.4 Hz to 0.7 Hz (Figure 4). Participants stood barefoot on the measurement platform, with arms crossed over their chest and head facing forward. Each measurement lasted 60 seconds, and the analysis focused on the period between the 15th and 45th seconds, during which oscillations occurred in tests with the oscillating scenes.

The measurement data were processed using MATLAB software. The analysis focused on COP displacements in the AP direction during EO and EC tests, as well as during the middle 30 seconds of tests with scene oscillations at 0.7 Hz and 1.4 Hz, including the 15 seconds before and 15 seconds after the frequency changes. Basic stabilographic parameters, such as COP velocity and COP displacement range in the AP direction, were calculated, along with the Trend Change Index (TCI). The TCI was expressed as both the total number of trend changes for the entire measurement and the number of trend changes within specific time intervals: 0–0.2 s, 0.2–0.5 s, and 0.5–1 s. Each time interval represents the time elapsed between consecutive trend changes. Additionally, the following values were calculated based on the TCI algorithm: the average distance between consecutive trend change points

(MACD_dS), the average time between consecutive trend change points (MACD_dT), and the average velocity of displacements between consecutive trend change points (MACD_dV).

For the statistical analysis, due to the non-normal distribution of the data, Friedman's ANOVA and Wilcoxon post hoc tests with Holm correction were applied. The analysis divided the results into three groups: standard time-domain values, TCI values for the entire measurement, and values calculated based on trend analysis (MACD_dT, MACD_dS, MACD_dV). COP velocity in the AP direction significantly increased after participants closed their eyes and when visual disturbances were introduced in VR. However, no statistically significant differences were found for TCI values between EO and EC. Statistically significant differences in TCI values were observed when comparing the EO test in the real environment with measurements in the virtual environment using the 0.7 Hz oscillating scene. Introducing VR disturbances decreased the median TCI values in the 0–0.2 s time interval. For the 0.2–0.5 s interval, higher median values were observed in the 1.4 Hz oscillating scene tests, while the highest values for the 0.5–1 s interval were observed in VR tests with 0.7 Hz disturbances. In terms of MACD_dS values, a significant increase was noted in VR tests compared to real-world measurements, but no significant differences were found between VR tests. For MACD_dT, significantly higher values were observed in measurements conducted at a 0.7 Hz frequency. Median MACD_dV values increased after participants closed their eyes and when virtual reality was introduced.

The results show that in most cases, the TCI value remained consistent across different testing conditions, suggesting that maintaining balance requires a certain number of postural corrections. The MACD_dS, MACD_dT, and MACD_dV values provided additional insights into COP movement, indicating whether changes in velocity result from shifts in COP distance, time taken, or both factors. The study also found that the MACD_dT value, which did not significantly decrease, plays a critical role in maintaining balance. The simultaneous increase in MACD_dS and decrease in MACD_dT—indicating longer COP jumps in shorter times—could lead to destabilization and potential falls.

Using stock market indicators to assess human stability complements standard time- and frequency-domain analyses. Analyzing TCI, MACD_dV, MACD_dT, and MACD_dS values offer new insights into factors affecting traditional balance parameters, such as path length, average velocity, and movement range. By integrating trend change analysis with stabilographic measurements, we can gain valuable information about the frequency of postural corrections, the intervals between corrections, and the speed of COP movement.

3.7 Trend change analysis in the assessment of body balance during posture adjustment in reaction to anterior-posterior ground perturbation [A5]

The fifth article in this series combines research on postural responses to external destabilizing stimuli [A2] with the innovative Trend Change Analysis (TCA) method for assessing balance [A3, A4]. Traditional methods of postural analysis, such as muscle activity measurements and center of pressure (COP) displacements, can be complemented by TCA. Inspired by stock market analysis techniques, TCA enables the identification of rapid postural corrections and the analysis of non-cyclical changes in the COP signal, providing new insights into balance maintenance strategies and responses to destabilizing stimuli. The article also highlights the need for further development and testing of TCA in real-world environments with actual balance-disturbing stimuli. This article hypothesizes that an increase in COP velocity may result from a shift in balance maintenance strategies, which should be reflected in trend analysis parameters, such as the number of trend changes and the time and distance between them. The aim of the study was to determine whether different conditions during the examination of postural preparation phenomena affect the values obtained through trend analysis.

The study involved 38 participants (27 women, 11 men), with an average age of 23 years, an average height of 172 cm, and an average weight of 70 kg. Exclusion criteria included previous lower limb injuries and balance issues. The testing setup consisted of a foot pressure measurement platform (WinFDM-S) and a treadmill (BalanceTutor) for postural perturbation training and assessment, which allowed for destabilizing ground displacements in the anterior-posterior (AP) and medio-lateral (ML) directions. A wireless electromyography system (Ultium EMG) and IMU sensors (Ultium Motion) were also used. The stabilographic platform was securely attached to the treadmill, and electrodes and EMG system sensors were placed near the muscle bellies of the tibialis anterior (TA), rectus femoris (RF), and the medial and lateral heads of the gastrocnemius (GM)—muscles actively involved in maintaining balance in the AP direction. An IMU sensor was attached to the treadmill belt, enabling detection of the onset of perturbations caused by platform movement and synchronization of all devices (Figure 5). The testing procedure involved three trials. In the first trial (Tr1), participants were unaware of the timing or direction of the perturbation. In the second trial (Tr2), participants knew the perturbation would occur but were unaware of its exact timing or direction. In the third trial (Tr3), participants were informed of the exact timing and direction of the perturbation, with a countdown displayed on a screen. In each trial, the platform made two movements - one forward and one backward - each 9.5 cm long and lasting 0.5 seconds.

Figure 5 Measurement stand [A5]

After the measurements were completed, both time-domain values and trend change analyses were conducted. The analysis included the calculation of the average COP velocity and several trend change characteristics: the total number of trend changes during the entire test (TCI), the average distance between consecutive trend change points (TCI_dS), the average time between consecutive trend change points (TCI_dT), and the average velocity of instantaneous displacements between consecutive trend changes (TCI_dV). The Shapiro-Wilk test revealed non-normal distributions for all variables, so statistical analysis was conducted using medians and non-parametric tests. The results from Tr1, Tr2, and Tr3 were compared using the Friedman test and Wilcoxon pair tests with Holm correction.

The EMG signal analysis showed differences in lower limb muscle activation across the trials, particularly in Tr3, where earlier preparation for the balance disturbance (EPA and APA) was observed [A2]. The studies confirmed that varying levels of muscle activation influenced COP signal characteristics, as reflected in changes to parameters such as TCI_dS and TCI_dV. Notably, in Tr3, a significant increase in the distance between consecutive trend change points was observed one second before the disturbance, suggesting a shift in balance strategy. This increase was also evident in the number of COP displacements within specific time intervals. When participants were aware of the impending disturbance, COP velocity (V_{COP}) increased, indicating a potential shift in balance strategy. Despite the increase in V_{COP}, the number of trend changes (TCI) remained constant, suggesting that these changes were due to longer distances covered by COP rather than more frequent postural adjustments. It was also observed that increased muscle tension, caused by the anticipation of the upcoming movement, led to body stiffening, which may have altered balance maintenance characteristics. These studies suggest that the body's responses to balance disturbances are complex and vary depending on the participant's awareness of the disturbance.

The trend change analysis method complements traditional COP analyses and EMG measurements in studies of postural preparation. TCI-based metrics help explain phenomena such as increased COP velocity, providing insights into the mechanisms underlying balance maintenance strategies.

3.8 Trend change analysis of postural balance in Parkinson's disease discriminates between medication state [A6]

The sixth article in the series describes the use of an innovative trend analysis method in balance studies utilizing IMU sensors in a group of individuals with Parkinson's disease (PD). These studies have the potential to significantly impact clinical practice by providing more precise tools for monitoring disease progression and developing new diagnostic methods through detailed postural stability analyses. The article was produced in collaboration with the University of Kiel, where the measurements were conducted.

The research presented in the article had two primary objectives: to investigate the potential application of trend change analysis (TCA) for assessing postural stability using IMU sensors and to apply this analysis in the context of neurological diseases, particularly PD. The hypothesis was that TCA could distinguish between individuals with Parkinson's disease (pwPD) and healthy adults, as well as differentiate between the "on" (PDon) and "off" (PDoff) phases associated with taking dopaminergic medications.

The study group consisted of 61 healthy individuals, divided into two subgroups: young adults (YO) comprising 40 participants with an average age of 30 years, an average height of 185 cm, and an average weight of 80 kg; and older adults (OP), comprising 21 participants with an average age of 73 years, an average height of 181 cm, and an average weight of 84 kg. Additionally, 29 individuals with Parkinson's disease (pwPD) participated in the study. Among the Parkinson's subgroup, 13 individuals were assessed as PDoff (UPDRS III score of 24 ± 10), 23 as PDon (UPDRS III score of 30 ± 20), and 7 were evaluated in both PDon (UPDRS III score of 26 ± 10) and PDoff (UPDRS III score of 27 ± 10) conditions. All participants with Parkinson's were inpatients at the neurogeriatric ward of the Neurology Center at University Hospital Schleswig-Holstein in Kiel.

Three IMU sensors (Ultium Motion) were attached to participants' bodies—on the pelvis, sternum, and head—using flexible straps (Figure 6). Participants were instructed to stand upright with their feet together and focus on a point on a white wall for 10 seconds as part of the Short Physical Performance Battery test.

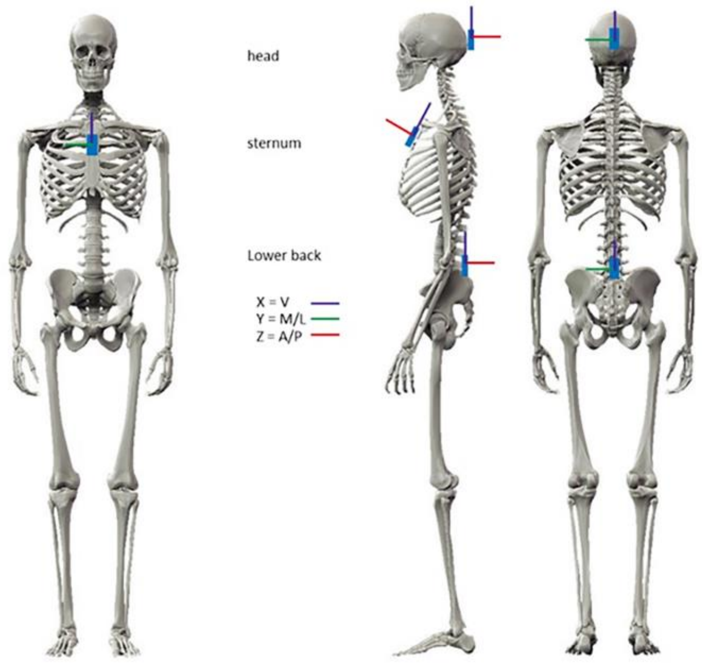


Figure 6 IMU sensors placement [A6]

Data from the IMU sensors were processed using MATLAB software. The following parameters were determined for center of mass (COM) movements: the sway jerkiness (JERK) (cm^2/s^5), the sway area (SURFACE) (cm^2), path (PATH) (cm), mean velocity (MV) (cm/s), acceleration range (RANGE) (cm/s^2), and root mean square of the acceleration (RMS) (cm/s^2). Additionally, trend change analysis (TCA) [A3, A4] was performed on the acceleration signal, yielding values for TCI (Trend Change Index), TCI_dT (average time between trend changes), TCI_dS (average distance between trend changes), and TCI_dV (average velocity between trend changes). Statistical analysis was conducted using MATLAB R2022a and JASP software. The Shapiro-Wilk test revealed non-normal distributions for all parameters, so non-parametric tests were applied. Differences between groups and sensor positions were analyzed using the Kruskal-Wallis test and post-hoc Dunn's test with Bonferroni correction.

The results confirmed that differences in movements between body segments can be detected using both traditional balance parameters and TCA. However, only TCA was able to distinguish between "on" and "off" states in individuals with PD. The findings suggest that during simple balance tasks, individuals with PD may exhibit behavior similar to healthy individuals, due to compensatory mechanisms in the central nervous system. The study showed that dopaminergic treatment influences postural stability. TCA results indicated a higher number of trend changes (TCI) and lower TCI_dT values in the "off" state compared to the "on" state, potentially reflecting visual and motor deficits caused by dopamine depletion.

The most significant conclusion from the study is that the introduction of trend change analysis (TCA) was crucial in detecting meaningful differences between the "on" and "off" treatment states in PD. This highlights TCA's potential for assessing disease-related changes that are not captured by conventional balance parameters.

3.9 Summary

The doctoral dissertation presents the results of research published in several scientific articles, focusing on the mechanisms of postural control and the ability to maintain human balance under various experimental conditions, including both real-world and virtual environments. The findings revealed that standard metrics, such as COP velocity - commonly used to assess balance - are insufficient to fully capture an individual's response to applied stimuli. Frequency analysis of head movements proved valuable, enabling the identification of individuals most susceptible to visual disturbances.

The obtained research results indicated that the standard time-domain measures used to describe balance ability are insufficient to determine the subject's response to applied stimuli. The application of frequency-domain analyses can significantly expand the interpretative possibilities of the acquired measurement data, as demonstrated in one of the studies assessing the impact of head movement on maintaining body balance. These analyses made it possible to identify individuals most susceptible to visual disturbances within the study group. In studies examining the influence of real-world destabilizing stimuli on postural preparation phenomena, it was confirmed that the key factor triggering postural preparation and readiness to respond to the disturbance was the information about the timing of the upcoming disturbance. Additionally, it was shown that combining stabilographic analysis, EMG, and IMU significantly enhances the ability to accurately assess balance maintenance mechanisms. The research results confirmed that the method of analyzing momentary postural corrections enriches standard time- and frequency-domain analyses by accounting for non-cyclical COP displacements. The usefulness of this method was demonstrated in both virtual environment studies with oscillating disturbances and real-world settings with actual destabilizing stimuli. The development of analyses based on detecting momentary postural corrections opens new opportunities for diagnosing balance disorders, particularly in relation to neurodegenerative diseases such as Parkinson's disease. The application of these tools in clinical practice enabled differentiation between the "on" and "off" states of dopaminergic treatment in PD. These results indicate the high potential of the applied method, including its use in the early diagnosis of postural stability changes that are not visible with traditional balance assessment methods.

In conclusion, the conducted studies successfully achieved the research objectives. The use of destabilizing stimuli, both real and virtual, significantly broadens the scope for analyzing balance capacity. The incorporation of methods that detect momentary postural adjustments, enhances the ability to interpret phenomena associated with body destabilization. This approach allows for more precise monitoring of postural stability, which is crucial for rehabilitation, fall prevention, and the diagnosis of neurodegenerative diseases.

4. Podsumowanie wkładu własnego

Wkład własny do niniejszej rozprawy doktorskiej został szczegółowo określony w oświadczeniach dotyczących indywidualnego wkładu procentowego i merytorycznego współautorów, które zostały podpisane przez pierwszego autora lub autora korespondencyjnego w przypadku każdej publikacji będącej częścią pracy zbiorowej.

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Wykaz skrótów

COP	*środek nacisku stóp / center of pressure*
COM	*środek masy / center of mass*
AP	*przednio-tylny / anterior-posterior*
ML	*przyśrodkowo-boczny / medial-lateral*
EPA	*wczesne przygotowanie posturalne / early postural adjustment*
APA	*antycypacyjne przygotowanie posturalne / anticipatory postural adjustment*
CPA	*kompensacyjne korekty posturalne / compensatory postural adjustment*
VR	*rzeczywistość wirtualna / virtual reality*
EO	*oczy otwarte / eyes opened*
EC	*oczy zamknięte / eyes closed*
BB	*badanie w wirtualnej rzeczywistości bez wizualnego bodźca ostrzegającego / measurement in VR without visual warning*
BZ	*badanie w wirtualnej rzeczywistości z wizualnym bodźcem ostrzegającym / measurement in VR with visual warning*
V_{COP}	*prędkość COP / COP velocity*
V_{head}	*prędkość przemieszczeń głowy / head velocity*
EA_{COP}	*pole elipsy COP / COP ellipse area*
DAD	*zakres ruchu głowy / head movement range*
A1h	*pierwsza amplituda harmoniczna / first harmonic amplitude*
FFT	*szybka transformata Fouriera / Fast Fourier Transform*
EMG	*elektromiografia / electromyography*
IMU	*czujnik bezwładnościowy / Inertial Measurement Units*
TA	*mięsień piszczelowy przedni / musculus tibialis anterior*
GM	*mięsień brzuchaty łydki – głowa przyśrodkowa / musculus gastrocnemius medialis*
GL	*mięsień brzuchaty łydki – głowa boczna / musculus gastrocnemius lateralis*
RF	*mięsień prosty uda / musculus rectus femoris*
MACD	*wskaźnik zbieżności/rozbieżności średnich ruchomych / Moving Average Convergence/Divergence*
TCI	*wskaźnik zmiany trendu / Trend Change Index*
PSD	*gęstość widmowa / power spectral density*
MACD_dT / TCI_dT	*średni czas między kolejnymi punktami zmiany trendu / mean time between subsequent points of the trend change*

MACD_dS / TCI_dS	*średnia odległość między kolejnymi punktami zmiany trendu / mean distance between subsequent points of the trend change*
MACD_dV / TCI_dV	*średnia prędkość zmian przemieszczeń między kolejnymi punktami zmiany trendu / mean velocity of changes of displacements between subsequent points of the trend change*
TCA	*analiza zmian trendu / trend change analysis*
PD	*choroba Parkinsona / Parkinson's disease*
pwPD	*pacjenci z chorobą Parkinsona / patients with Parkinson's disease*
PDon	*pacjenci z chorobą Parkinsona przyjmujący leki dopaminergiczne / patients with Parkinson's disease during medication on*
PDoff	*pacjenci z chorobą Parkinsona nieprzyjmujący leków dopaminergicznych / patients with Parkinson's disease during medication off*
JERK	*zryw / sway jerkiness*
SURFACE	*pole elipsy podparcia / sway area*
PATH	*ścieżka / path*
MV	*średnia prędkość / mean velocity*
RANGE	*zakres przyspieszenia / range of acceleration*
RMS	*błąd średniokwadratowy/ root mean square of the acceleration*

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Analysis of center of pressure displacements and head movements triggered by a visual stimulus created using the virtual reality technology

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Purpose: The purpose of the study was to determine how a stimulus presented in the virtual reality environment as a simulation of a fall off the stairs, triggers a loss of balance. The study also examined if the head movement measurements and the analysis in the frequency domain could increase the range of interpretation. *Methods*: 11 healthy individuals were tested, two [A1] were identified as more susceptible to the introduced disturbance, and one reported having dizziness, car sickness and fear of heights. Measurements of center of pressure (COP) and head positions were performed in the real and in the virtual environment. The beginning of the simulation was either unexpected or preceded by a signal. The analysis included standard parameters determined in time domain as well as the amplitude of the first harmonic from the fast Fourier transform (FFT). *Results*: The analysis did not reveal statistically significant differences between results obtained: in real and virtual environments, with and without the warning signal. It was possible to notice the effect of virtual disturbance in the three selected individuals; this was particularly evident in the analysis of the first harmonic of the FFT. *Conclusions*: The conducted tests revealed that the limitation of the analyses exclusively to the time domain could be insufficient for a comprehensive interpretation. The effect of introduced disturbance was particularly noticeable in the analysis of the first harmonic for head movement. The application of this parameter could enable a more accurate investigation of a strategy aimed at maintaining balance.

Key words: *virtual reality, postural stability, balance, frequency analysis*

1. Introduction

The ability to maintain balance is important in enabling independent life, the performance of everyday activities and in allowing individuals to work at heights or partake in competitive sports. Disturbed balance may indicate diseases of the nervous system or be the result of ageing. Difficulty in maintaining balance could lead to falls resulting in injuries, which may further worsen health conditions [4], [22]. It is estimated that falls may impact 30% of people older than 65 who live independently at home and more than 50% of people in retirement homes [17]. Since society is aging, the number of people at risk of falls could be expected to grow [11]. However, the risk of falling does not only affect the elderly and has particularly serious consequences for people working at heights. A fall from height is one of the most frequent reasons for injuries (48%) and death (30%) among all incidents occurring during work at heights [23]. Apart from irresponsible behavior and inexperience, an increased risk of falling from a height is connected with age, fatigue, insufficient sleep, and poor health [23]. All of these factors adversely affect concentration and may lead to disturbed balance, increasing the risk of falling.

Problems with balance can be permanent and can be manifested by noting changes in the values of parameters that indicate the ability to maintain balance. These can be verified in tests involving regular standing with eyes open or closed, as well as during

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gait [2], [3], [10], [31], [33]. However, there are cases when problems with balance are caused by specific conditions in the environment or from specific behaviors. For instance, American football players developed problems with balance after quick right- and left-head movements [7]. The detection of such less obvious balance disturbances requires the extension of standard balance measurements to include simulations of events connected with work or the introduction of additional factors that could disturb balance. The information delivered to individual senses responsible for balance may be diversified during a conflict of sensory stimuli. This is why balance-related tests are increasingly performed using virtual reality technology, where the scenery and the manner of its presentation are used as balance disturbing stimuli [6], [9], [16], [19], [20].

The application of virtual reality makes it possible to simulate the visual conditions of balance loss or to trigger a balance destabilizing factor. Tests of visual stimulation and the effect of visual stimuli on posture control were performed under various conditions. These included individuals focusing on objects located at various distances [28] or being exposed to a moving environment [16], [25], [32]. Virtual reality (VR) creates an illusion where a test participant has the impression of being in a place other than where they really are [18]. The introduction of virtual surroundings while simultaneously maintaining immovable ground, exposes a person to sensory conflict. As a result, the body generates a kinetic response towards the visual disturbance. The analysis of such a response can be a valuable source of diagnostic information concerning disturbed balance [9], [20].

In tests of maintaining balance, analyzed parameters usually include center of pressure (COP) displacements, including the average COP velocity, the area of an ellipse, and the range of the COP movements in the anterior-posterior (AP) and mid-lateral (ML) directions [5], [12], [14]. Importantly, there are increases in the values of these parameters under conditions of conflicting sensory stimuli, however, these changes are not necessarily connected with disturbed balance [15], [16]. For this reason, there is a growing demand for new methods of measurements and analyses. One of the solutions involves completing a frequency analysis aimed at observing changes that may be unnoticeable during a standard analysis using a time domain [12], [16]. Force platform-related analyses are limited to the two-dimensional plane under feet. This is why some tests have involved the use of accelerometers to detect changes in the postural balance based on movements of the center of mass (COM) or other segments of the body [20], [21], [34]. This resulted in additional analysis of body movements for maintaining balance [24], [29].

1.1. Objective of the research

It seems necessary to apply extended methods and use an advanced mathematical apparatus for measuring the ability to maintain balance. However, this approach requires that there would be standardized procedures for such measurements and defined ranges of parameter values, taking the responses of healthy and ill individuals to introduced disturbance into account. For this reason, the purpose of the research work included the following:

- determination whether a simulated fall off the stairs may affect the posture,
- determination of whether the extension of analyses with a frequency-related analysis could increase the scope of interpretation in relation to analyses based on the time domain,
- determination of whether measurements of the head movements could supplement the COP measurements with information regarding the effect of the introduced disturbance factor on the ability to maintain balance,
- determination of whether the visual warning signal preceding the introduction of the balance-disturbing factor will trigger a change in the head movements and the COP displacement.

2. Materials and methods

2.1. Study group

The study group consisted of 10 test participants (7 females and 3 males) at an average age of 25 (SD = 3) and an average BMI of 23 kg/m^2 (SD = 3.3). Interviews of participants indicated that none had a history of an extreme lower limb injury or had any motor system dysfunction, balance disorder, or any other problem that could affect their balance.

An additional test participant, designated as pp3 (female, 25 years of age, BMI = 17.3 kg/m^2), declared that she suffered from intense car sickness, a fear of heights, and dizziness (when her eyes were closed).

Increased values of selected parameters for two individuals (designated as pp1 and pp2) suggested the presence of potential problems with maintaining balance or, at least, higher susceptibility to the introduced virtual disturbance. Therefore, these individuals were excluded from the group of individuals treated as healthy and were instead the subjects of a separate case study, along with the person suffering from balance maintaining problems (pp3).

This study was previously approved by the Ethics in Research Committee of the Academy of Physical Education in Katowice (protocol number 5/2020). Each of the study participants was informed of consent in accordance with the Ethics in Research Committee, as well as of the form and the course of the study. In each case, consent to participate in the study was expressed in oral form (written consent was not required).

2.2. Measurement stand

The study was performed using a measurement platform (WinFDM-S, Zebris, sample frequency 100 Hz, 2560 extensometer sensors, sensors area: 34 cm × 54 cm), and a VR HTC Vive headset. The VR application was prepared in the Unity 3D system and consisted of a simple scenery. In the application, the avatar (facing the stairs) was placed on the floor near the stairs. The scenery is presented in Fig. 1.

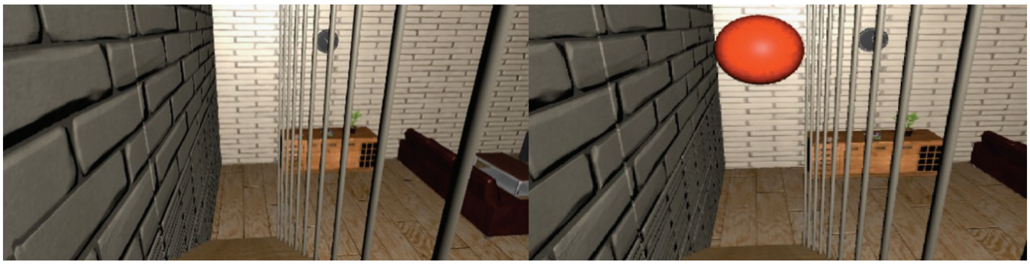

Fig. 1. The scenery of the test: without and with the warning signal

2.3. Experimental procedure

Measurements were taken of successive positions of the center of pressure (COP) (WinFDM-S) and successive positions of the head (HTC Vive). For head movement, the current position of the head was read directly from the Unity graphics engine from data obtained from the HTC Vive system. The results were used for a frequency analysis; the average COP velocity and the area of the ellipse containing the COP, were calculated.

The performance of the tests in the virtual reality trial was preceded by a trial of standing on the platform, where participants had their eyes open (EO) and closed (EC) for 60 seconds. Afterward, the participants took part in tests involving virtual reality (BB). The first 20 seconds of the tests involved leaning in the ML direction to synchronize the measurement systems. From the 20th second onwards, the patients stood still. At the 30th second, the system showed the movement of the scenery, simulating the fall off the stairs. For 20 seconds following the disturbance, the subject was supposed to stand without moving. In the subsequent test (BZ), the measurement procedure was the same as BB, except that participants were exposed to a visual red signal three seconds before the disturbance was triggered. The tests were subsequently repeated three times, obtaining data designated as BB1, BZ1, BB2, BZ2, BB3, and BZ3. During each of the measurements, the participant was supposed to stand motionless with their arms crossed on their chest.

2.4. Analysis of results

The positions of the COP and the head on the platform plane and in space were recorded. The analyses included parameters identified as changes in the direction of the triggered visual disturbance (i.e., AP). Displacements in the ML direction were only used to synchronize related courses and were not analyzed. The values recorded during the first 20 seconds were not analyzed and were only used for time synchronization of the results. The analysis involved measuring the changes of the COP displacements and head movements and adopting the 20th second (the end of the synchronizing procedure) as the zero moment of the test.

The analysis of the obtained measurement data was divided into three stages (Fig. 2).

The first stage involved identifying the effect of disturbance based on comparison of the average velocity of the COP displacements and head movements (V), and the area of the ellipse (EA) in relation to the

COP and the head. The comparison of the test results included the EO, EC, BB1, BB2, BB3 and BZ1, BZ2, BZ3 tests.

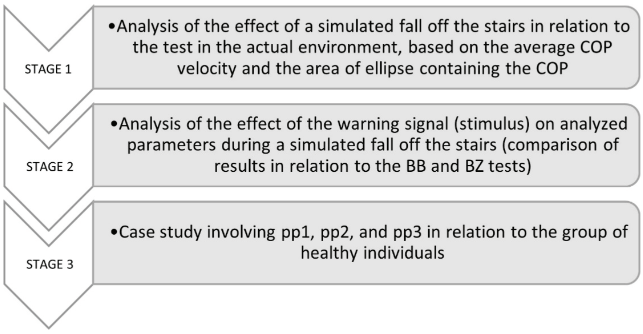

Fig. 2. Stages of analysis of the results

The second stage involved the analysis of the COP displacements and head movements in the AP direction in the time domain and frequency domain. The aim of the analysis was to identify the impact of the warning signal (stimulus) on the test participants' responses to the disturbance. The analysis included the average velocity, the area of the ellipse, and the range of movements in the AP direction, for both the COP and the head. The range of the COP and head movements was calculated as the difference of the coordinates of two extreme positions in the AP direction, within 15 seconds following the introduction of disturbance (recorded after a three second disturbance). The displacement values were designated as D_{AD}. The COP displacement and head movement values in the AP direction were used for the frequency analysis by calculating the fast Fourier transform. The first harmonic amplitude (A1h) was identified from an analysis of the spectrum. Supplementing the analysis in the time domain with the frequency domain enabled examination of the periodicity of the movements.

It was necessary to compare values to establish whether the following results: V, EA, D_{AD}, and A1h, changed in successive trials (three repeated tests in relation to BB and BZ). The results revealed statistically nonsignificant differences between the values of the parameters in subsequent tests. This result led to a decision to combine values of the BB1, BB2, and BB3 tests into one BB group as well as to combine values of the BZ1, BZ2, and BZ3 tests into one BZ group. The lack of statistically significant differences in subsequent measurements (all of the obtained *p*-values from the Kruskal–Wallis test were higher than a confidence threshold of 0.05 in all of the tests) made it possible to group all of the measurement results into two sets (BB and BZ) and perform further analyses taking into account the increased size of the groups.

An analysis of differences between the range of the COP displacements and the range of head movements triggered by the visual disturbance was completed. This was done by comparing parameter values after the introduction of disturbance, i.e., D_{AD}, and A1h for the COP and the head, for each of the tests.

The analysis made it possible to establish whether the visual warning signal/stimulus affected the behavior of test participants. The Shapiro–Wilk test was used for the BB and BZ groups and to perform comparative tests. This non-parametric test was conducted because of the small group size and the lack of a normal distribution. All calculations were performed using the Statistica, version 13, software program.

The third stage of analysis encompassed a case study comparing parameters for the individuals potentially having difficulty maintaining balance with those of the persons declaring a lack of balance-related problems.

3. Results

The first stage of the analyses used measurements in the real environment and involved the calculation of the average velocity (V) of the COP displacements and the values of the ellipse area (EA) containing the COP. The test results are presented in Table 1.

Table 1. Average COP velocity and the area of ellipse obtained in relation to the EO and EC tests (in relation to the pp1, pp2, and pp3 tests; the values indicate the median of 3 measurements)

	V [mm/s]		Ellipse area COP [mm^2]	
	EO	EC	EO	EC
Median	11.5	11.6	107	151
Lower quantile	9.3	10.4	76	80
Upper quantile	12.8	14.0	172	196
Minimum	7.9	8.1	67	40
Maximum	14.0	19.8	558	696
pp1	11.2	12.4	406	661
pp2	9.9	9.2	98	107
pp3	16.1	11.1	230	567

The average velocity of the COP and the head, the ellipse area containing the COP and head positions, and the ranges of the COP and head movements in the AP direction (D_{AD}), measured in the virtual environment were calculated (Table 2). The amplitude of the first harmonic in the fast Fourier transform (FFT) spectra was also calculated based on the frequency

Table 2. Values of velocity (V), range of movements in the AP DAD direction, the ellipse area calculated based on the ranges of the COP displacements, and head movements for the BB and BZ tests (the values indicate the median of 3 measurements in relation to the pp1, pp2, and pp3 tests)

	V 25s from the loss of balance				D_{AD} [mm]				Ellipse area [mm^2]			
	BB		BZ		BB		BZ		BB		BZ	
	COP	Head	COP	Head	COP	Head	COP	Head	COP	Head	COP	Head
Median	12.5	8.7	14.0	8.3	26.8	34.4	23.4	35.9	149	342	148	305
Lower quantile	11.5	8.0	11.7	7.9	18.6	25.8	19.2	27.7	104	242	93	258
Upper quantile	16.0	9.7	14.6	9.2	32.9	46.9	30.8	41.2	230	522	199	469
Minimum	9.2	6.4	9.5	6.8	14.2	19.8	15.7	17.5	54	174	47	159
Maximum	23.6	12.2	20.7	12.5	60.0	82.0	41.5	56.3	487	949	444	891
pp1	11.8	10.7	17.7	15.3	34.4	56.9	64.2	94.7	259	898	249	815
	15.4	13.1	14.0	11.5	37.6	62.1	46.8	75.7	287	893	535	1317
	14.1	12.0	13.1	11.4	49.4	71.9	58.2	109.5	309	1433	388	1505
pp2	12.9	15.3	13.4	9.4	25.8	54.2	18.4	41.4	231	2459	335	1049
	13.5	11.4	11.1	9.2	27.1	47.9	22.9	26.7	623	841	210	553
	20.9	11.7	12.8	10.8	46.1	50.1	28.8	76.7	873	1514	521	2506
pp3	18.8	11.7	19.2	12.8	31.5	47.3	54.2	50.7	453	971	1450	1687
	13.5	9.2	16.8	11.1	32.5	54.9	35.3	64.3	277	530	628	1282
	16.2	10.3	18.5	12.1	22.1	49.6	25.1	38.3	386	733	674	848

domain; this was obtained from the changes of the COP and the head in relation to the tests performed in the real environment. The values are presented in Table 3.

Table 3. Values of the amplitude of the first harmonic A1h and the A1h ratio calculated for the head in relation to A1h calculated for the COP for the BB and BZ tests

	A1h [mm]				Head_A1h / COP_A1h	
	BB		BZ			
	COP	Head	COP	Head	BB	BZ
Median	10.2	17.2	9.5	15.3	1.5	1.7
Lower quantile	8.8	13.9	7.4	12.8	1.4	1.5
Upper quantile	12.6	18.5	13.4	22.4	1.8	1.8
Minimum	4.9	6.8	4.2	7.6	0.7	1.3
Maximum	14.7	28.0	16.6	28.1	2.9	2.6
pp1	29.1	51.6	31.9	58.1	1.8	1.8
	15.3	32.7	25.6	48.7	2.1	1.9
	36.6	62.2	32.4	72.8	1.7	2.2
pp2	12.1	38.5	11.4	31.6	3.2	2.8
	11.9	41.0	13.9	15.9	3.4	1.1
	17.0	58.1	24.6	85.9	3.4	3.5
pp3	18.2	37.6	12.5	29.7	2.1	2.4
	22.5	43.7	14.5	38.3	1.9	2.6
	13.0	31.0	12.0	18.5	2.4	1.5

The values of V and EA in relation to the EO, EC, BB and BZ tests revealed statistically nonsignificant differences (in relation to *V*: ANOVA KW $p = 0.26$; in relation to the ellipse area (EA): ANOVA KW $p = 0.59$).

The second stage of analyses involved testing for differences between the calculated values of V, EA, D_{AD} (Table 2) and A1h (Table 3) in relation to the tests with (BZ) and without (BB) the signal warning disturbance. There were no statistically significant differences based on the Wilcoxon signed-rank test ($p \gg 0.05$).

The values of D_{AD} and A1h (Tables 2 and 3) were compared to examine differences between the displacement of the COP and the movement of the head. This analysis indicated the existence of significant differences when comparing D_{AD} and A1h for the group of healthy individuals ($p \ll 0.05$ for the Wilcoxon signed-rank test).

The third stage of analyses involved examining how the results for the individuals who qualified as more susceptible to the introduced disturbance could differ from those in the control group. The V values for pp1 and pp2 (all measurements) did not deviate from the medians obtained for the control group. The velocity was higher only for pp3. However, there were differences in the ellipse area (EA), with increased values for pp1 and pp3 for all measurements. There were increased values of EA for pp2 but only for the measurements in the virtual environment. The maximum and minimum values for the group of healthy individuals were single events, i.e., they appeared for a given person in one measurement, whereas values of the remaining measurements were similar to the median. As a result, the single high value data items could be treated as accidental losses of balance that were without diagnostic significance.

4. Discussion

Analyses needed to be completed in two directions to investigate how a simulated fall off stairs could trigger a loss of balance and whether an analysis of COP displacements and head movements could enable the identification of balance-related problems [34]. The first direction analyzed changes in parameters describing destabilization and postural compensation for the group of healthy individuals. Similarly to the research by Huweler [13], the analysis focused on head movements and COP displacements, and obtained displacement values in the AP direction. Related comparisons were completed in the real and virtual environments. To assess the effect of the visual stimulus (signal) on the COP displacements and head movements, it was necessary to analyze the average velocity, the ellipse area of prediction, and the range of movements after the disturbance. Similarly to the tests by Alpers [1] and Davis [8], the analysis in the frequency domain was performed to determine the amplitude of the first cyclic component of A1h.

The second direction compared results for individuals diagnosed as experiencing an impact of the introduced disturbance on balance with those for individuals who were healthy.

4.1. Analysis of the impact of the scenery disturbance on healthy participants

4.1.1. Comparison of the measurements in the real and virtual environments

The first stage of analyses included the comparison of the most frequently measured stabilometric parameters [6], namely, the area of ellipse (which made up 95% of the path of support) and the average velocity of the COP displacements. The comparison was performed on the measurements in the real environment with the participants having their eyes open and closed (Table 1), and in the virtual environment using the additional stimulus triggering the loss of balance (i.e., simulated fall off the stairs) (Table 2). The significant power of the test (>0.8) combined with statistically nonsignificant differences between the results of the measurements indicated that the introduced destabilizing stimulus did not affect the behavior of the test participants. It is possible that the classical analysis provided insufficient information regarding if, and how, the introduced visual stimulus affected changes in postural stability [16], [33]. The results and interpretative ambiguity should be unaffected by the small group size because effective statistical analysis in relation to small groups in similar postural stability-related tests has previously been well demonstrated by Keshner et al. [19]. The primary factor of importance might be that data interpreted in the aforesaid manner could be sensitive to disturbance (if any) in the form of losses of balance unrelated to the ability to maintain balance. This seems to be confirmed by large discrepancies in the values of such parameters in different research studies. For instance, in the tests performed by Błaszczyk et al. [2] on a group of approximately 21 year-olds (the use of force-plate posturography), the average velocity of the COP amounted to 9.8 mm/s (SD = 1.6) with eyes open and 12.2 mm/s (SD = 2.7) with eyes closed. In the tests performed by Skalska et al. [27], the average velocity of the COP amounted to 5.9 mm/s (SD = 2.5) with eyes open and 6.0 mm/s (SD = 2.2) with eyes closed. In previous tests by other authors, the average velocity of the COP amounted to 8.1 mm/s (SD = 0.8) with eyes open and 10.5 mm/s (SD = 2.3) with eyes closed. An even greater divergence could be observed for the ellipse area. In two independent tests performed by Błaszczyk et al. [2] and Rocchi et al. [26], the values of the ellipse area for persons aged 60–70, were 179.5 mm^2 (SD = 114.1) and 484 mm^2 (range of 197 mm^2 to 998 mm^2), respectively. Furthermore, previous tests on a group of individuals who were approximately 21 years of age provided different values, i.e., 154 mm^2 (SD = 94.3) with eyes open and 174 mm^2 (SD = 104.7) with eyes closed [15]. Such differences and the standard deviations of the COP-related ellipse area (for which the relative error often exceeded 50% of the average) could indicate that the analysis is insufficient (particularly under conditions of conflicting sensory stimuli). Therefore, it was necessary to extend the above analyses with parameters to enable a better and more comprehensive interpretation of findings.

4.1.2. Analysis of the impact of the warning stimulus

The use of the signal warning about the simulated fall off the stairs made it possible for a test participant to prepare for the situation [34]. This was done to determine if preparation for an expected disturbance would affect the behavior of test participants; to achieve this it was necessary to analyze parameters identified after the disturbance, i.e., D_{AD} and A1h (Tables 2 and 3). The HTC VIVE system displaying VR images was used to identify successive positions of the head (specifically, the VR headset in space). In this way, it was possible to extend analyses based on the COP measurements with analyses based on head movements.

There were no differences in the parameters with and without the use of the warning signal. Such differences are frequently noticeable in cases where actual stimuli are triggering the loss of balance [1], [5], and when the preparation (e.g., an impact) could lead to a delayed reaction. In terms of the tests, the disturbance in the form of a simulated fall off the stairs was purely visual and its effect was probably so strong that even the information about its appearance and previous experience of what it would look like did not change the response of participants.

In addition, the analysis of all obtained results indicated greater head movements. Taking into consideration the movement of the entire body as that of an upended pendulum, the above phenomenon was natural and confirmed by results from Wider [30].

The measurements made it possible to identify three individuals where there was a significant effect of the introduced disturbance on the ability to maintain balance. These three individuals were the subject of a separate case study.

4.2. Case study

During measurements, it was possible to single out three individuals (one based on the previous declaration and two based on the measurement results), where the values of selected parameters differed significantly from those of the remaining individuals. The recorded differences could indicate problems with maintaining balance when events affect the perception of the environment (e.g., being at height or an unexpected movement of the ground) [20], [34]. Particular attention should be paid to the fact that the greatest differences were demonstrated in parameters rarely used in such measurements (i.e., the cyclic components of the head movements determined using the fast Fourier transform).

4.2.1. Analysis of average velocity and the ellipse area of prediction

There were no clear differences between the group of healthy individuals and the three diagnosed with increased susceptibility to the introduced disturbance when examining the average velocity of the COP and the head (Table 2) (for measurements taken in the real environment). Although these parameters for the three individuals were higher than the median for the group, they were lower than the maximum values for the group. Thus, the parameters did not explicitly differentiate those individuals who were strongly affected by visual disturbance; this indicated that further analyses were needed.

Differences were noticeable when viewing the COP-related ellipse area (Table 2). All of these values for the three individuals (during measurements in the virtual environment) were higher than the upper quantile. In addition, some of the measured values exceeded the maximum values for the group (for pp1 – one measurement, and for pp2 and pp3 – three measurements). Furthermore, the ellipse area values (which were identified for the head movements) for the three cases (Table 2) were higher than the upper quantile of the reference group. In half of the cases, these values were also higher than the maximum values for the healthy participants. The increased values of the ellipse area, particularly those obtained with the head movements, could indicate that a given participant found it difficult to maintain balance after experiencing the disturbance. However, such differences were only observed for some measurements, which could imply a momentary loss of balance.

The range of the COP displacements and head movements (Table 2) did not unequivocally indicate the scale of the disturbance effect on the three individuals. Only pp1 revealed increased values of the analyzed parameter. In half of the cases, the values were higher than the maximum values for the group. For the remaining cases, the values were higher than the upper quantile. For pp2 and pp3, the values varied from similar to the median to sporadically higher than the maximum for the group.

These ambiguities could limit the usefulness of such measurements in diagnosing problems with maintaining balance. For this reason, it is necessary to find parameters which could provide the most repeatable results.

4.2.2. Analysis of the first harmonic – balance maintaining strategy

The frequency analysis of the tests concerning balance made it possible to break down the signal composed of changes in the positions of the COP and those of the head into successive cyclic components, giving the frequency and amplitude of such movements [15], [16], [19]. The use of the fast Fourier transform in the tests revealed that the application of the destabilizing stimulus (in the form of a simulated fall off the stairs) triggered the appearance of the cyclic component with the highest amplitude of 0.1 Hz to 0.2 Hz. The analysis of this for the reference group and pp1, pp2, and pp3 (Table 3) did not reveal differences between the tests with and without the warning stimulus. However, there were significant differences for the three individuals (i.e., pp1, pp2, and pp3), compared to the remaining participants. These differences were visible early when comparing the values of the amplitude of the first harmonic obtained with the COP displacements, yet the greatest differences were observed with this parameter in relation to the head movements. In the second case, only two values (i.e., one for pp2 and one for pp3) were lower than the upper quantile of the group. The remaining values were between 1.3 and 3.8 times higher than the upper quantile and between 1.05 and 3.1 times higher than the maximum values for the rest of the group. Such repeatability of increased amplitude of the harmonic peak was not observed for any other test participant. Such increased values obtained for other persons were sporadic results, obtained in single measurements for a given person.

The above results indicate that, when analyzing the effect of disturbances due to virtual reality and, probably in other balance-related tests, it is necessary to take the COP displacements (which is a certain standard) and movements of other parts of the body (e.g., head – as in the case of this study) into consideration. Even when the head movements are taken into account, an ordinary analysis in the time domain could be insufficient for detecting differences among test subjects. For the tests discussed in this article, only separate analyses of individual cyclic components indicated differences that distinguished the three individuals from the entire group.

Furthermore, the determination of the first harmonic indicated the strategy adopted by individual participants to maintain their balance. The proportion of the first harmonic amplitude for the head and the COP (Table 3) indicated whether a given person balanced using their entire (stiffened) body (value of the ratio close to 1) or whether head movements dominated over the COP displacements. The latter case, as could be surmised (because it was not directly measured) involved a larger range of trunk movements combined with the simultaneous flexion/extension of the hip joint and dorsal/plantar flexion of the ankle. The appropriate synchronization of movements causes an increased range of trunk movements and a slight range in the COP displacements (all in the AP direction). The tests revealed that the aforementioned strategy dominated in the three individuals as indicated by increased parameter values. This implies that individuals having difficulty maintaining balance will apply the above strategy of balancing their body in virtually disturbed (i.e., moving) surroundings. This also explains slight differences in parameters identified on the basis of the successive COP displacements. The above conclusion could be important in diagnosing problems with maintaining balance, but requires more tests and verification.

The determination of a coefficient describing the strategy for maintaining balance should not only include parameters determined in the time domain (e.g., ranges of the COP displacements and head movements), but should primarily include the harmonic of the FFT that is characterized by the highest amplitude. Such an analysis could help filter out slight movements, which do not necessarily determine the global movement strategy.

5. Conclusions

The application of the VR technology along with the creation of visual stimuli aimed to trigger the loss of balance could help diagnose balance-related problems and extend tests of balance involving standing with one's eyes open and closed. The VR-aided simulation of phenomena, which could occur in reality, enables the activation of mechanisms for achieving an appropriate posture following exposure to disturbances. The proper understanding of these underlying mechanisms could help in the development of systems and methodology for detecting balance-related problems. Such problems are manifested by an improper reaction to the upsetting of the balance of the stabile posture; these may be undetectable during tests in the real environment, which is why VR technology can be helpful.

The results revealed that measurements under conditions of conflicting sensory stimuli required an extension of the traditional approach for a more comprehensive analysis of results. The parameters most commonly analyzed when assessing the ability to maintain balance do not provide a full answer as to how the stimuli introduced to upset balance affects a given person. Thus, this study recommends the use of analyses in the frequency domain to provide a more comprehensive view of how upsetting the balance impacts individuals. The tests also revealed that measurements of the successive COP positions (currently the most frequent method of assessing the ability to maintain balance) could prove insufficient in analyzing balance. Additional measurements with head movements extended the possibilities of interpreting results obtained in related balance tests.

The amplitude of the first harmonic, particularly with respect to the head movements, most repeatedly differentiated the individuals with the highest susceptibility to the introduced disturbance. It should be emphasized that such results do not necessarily indicate the existence of health disorders; however, the increased susceptibility to certain forms of disturbance could translate into unexpected behavior during work at heights or when playing sports requiring fast movements of the head.

Conflict of interest

The authors declare no conflict of interest. The authors declare that all authors were fully involved in the study and the preparation of the manuscript and that the material within has not been and will not be submitted for publication elsewhere.

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The effect of selected lower limb muscle activities on a level of imbalance in reaction on anterior-posterior ground perturbation

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Purpose: We investigated whether an increase in muscular tone induced by the information about imminent posture destabilisation brings a positive result and prevents such destabilisation. *Methods*: We measured forward and backwards movements of 38 participates (27 females and 11 males, aged 23 (SD 2.6)) on the treadmill (forward and backward movements). All participants were subjected to three test condition trials (Tr): 1) subject did not know the nature and time of perturbation (Tr1); 2) subject knew the nature of perturbation but did not know time (Tr2); 3) both the time and nature of perturbation were known precisely (Tr3). The tests resulted in the determination of muscular activity connected with a postural adjustment as well as values of pressure exerted by the forefoot on the ground, and the angle of flexion in the knee joint. *Results*: In terms of postural adjustments, it was possible to observe statistically significant differences in muscular activity between Tr1 and Tr2 with reference to Tr3. No statistically significant differences were identified in all phases regarding values of forefoot pressure and those concerning the angle of flexion in the knee joint. An increase in the muscle tone before perturbation was correlated with the displacement and the velocity of the COP after perturbation. *Conclusions*: The results obtained indicate that knowledge of the expected time of perturbation is responsible for postural adjustment. Furthermore, muscle tone resulting from an adjustment of perturbation and responsible for the stiffening of lower limbs triggered greater displacement of the COP after perturbation.

Key words: *muscle activation, disturbance response, postural stabilization, postural compensation, anticipatory postural control*

1. Introduction

Mechanisms of Anticipatory Postural Adjustment (APA) and mechanisms of Early Postural Adjustment (EPA) are methods enabling adjustments of the body in response to a given stimulus [1], [7], [25], [39]. The above-named adjustments may take various forms, usually affecting movements of the entire body, or its parts. APAs are manifested in a variety of manners, yet many researchers primarily indicate changes of muscular activity [25] or changes concerning the displacement of individual segments of the body, the centre of mass [3], [4], or the centre of pressure [1], [7]. Research performed by Cleworth et al. [7] and Xie et al. [41] revealed that postural adjustments range from changes preceding the performance of a given movement to the moment of a bodily response, as well as subsequent stimulus-evoked reactions. There are two types of stimuli that relate to foregoing. One is connected with the internal initiation of a movement (movement initiated by someone's own intention) as indicated by Shumway-Cook and Woollacott [30] as well as Xie et al. [41] and the other stimulus, con-

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nected with an external factor (frequently leading to postural destabilisation) provoking a bodily movement as indicated by Cleworth et al. [7], Sibley and Etnier [31]. Initially, the researchers focused on the first stimulus type, reducing analyses of APAs entirely to the processes taking place in the body and those followed by the movement. However, recent research suggests that the transfer of energy from an external source (outside the body) can also trigger postural adjustment if it is only known when the expected energy stimulus takes place. For instance, Mohapatra and Krishnan [25] reported a lack of APAs in relation to random moments, when test participants were not informed about possible perturbation and the permanent occurrence of APAs when such information was available. It was also possible to demonstrate the existence of correlation between APAs and the value of force applied from outside. In their research, Berg and Hughes [2] found that APAs were greater when the test participant was not aware of the magnitude of the aforesaid force. The lack of information about the nature and form of perturbation resulted in an increased muscular activity.

APAs have a practical application of discovering postural disorders and the prediction of falls [1], [26], [32], [36]. Ritzmann et al. [36] indicated the practical use of such research in the development of training aimed to reduce fall-triggered injuries. Additionally, the observation of APAs can be utilized during the detection of limb stability and the identification of recovery (e.g., after injuries or surgeries). Oeffinger et al. [26] pointed out to the practical use of APA measurements in the assessment of recovery after ligament reconstruction. In their research, Scariot et al. [29] attempted to correlate APAs and EPAs (taking place slightly earlier) with perturbation reactions or compensatory postural adjustments (CPAs). It can be concluded that the above-named tests were an attempt to answer the question whether the occurrence of APAs positively affected the prevention of destabilisation triggered by external factors. There are multiple interpretations of Scariot and colleague's [29] results. However, when focusing on postural stability, the result could be interpreted in a way similar to that developed in the research by Bax et al. [1], which suggested that the smaller displacement of the COP (center of pressure) and the COM (center of mass) after perturbation reflects greater efficacy in maintaining postural stability. The above-presented assumption is incomplete as it does not include a number of forces and dynamic parameters compensating numerous phenomena which take place in the human locomotor system and whose aim is to restore the bodily posture preceding perturbation [15], [20], [21], [38]. However, the COM and COP displacements are measurable and reflect the correlation with the probability of falling (which was demonstrated in related research) [13], [14], [19], [41], [43].

Inspired by the research described previously, the work described below seeks to determine whether information concerning the time of ground-related perturbation affects the muscular tension of lower limb muscles before the perturbation. An attempt to answer the above-presented question requires the identification of the occurrence of APAs and EPAs. It should also be determined whether increased muscular activity was continuous over a given time or was abrupt and shortly preceded perturbation. Based on these findings, we next sought to determine whether lower limb muscle tone at the initial EPA-related stage leads to the increased muscle tone in the phase connected with APA.

It is also worth considering that muscular activity is not the only manifestation of postural adjustment, research-related tests should also include changes of the COP as well as postural changes connected with anticipatory ("pre-perturbation") adjustments. Another research question is whether an increase in the muscular tension of lower limb muscles before perturbation (i.e., an increase in APAs) results from postural changes triggering the displacement of the COM forwards or backwards, and, consequently, an increase or decrease of pressure exerted by the forefoot on the ground. The final question is whether an increase in the muscular tension of lower limbs a moment before perturbation (an increase in APAs) results from postural changes connected with an increase in the angle of flexion in the knee joint.

The occurrence of APAs entails certain consequences connected with the manner of a response to perturbation (identifiable by measuring some parameters after the occurrence of perturbation). Therefore, it should be considered whether an increase in the muscle tone preceding perturbation (an increase in APAs) leads to an increase in the COP velocity and movement range after perturbation. Finding the answers to the questions formulated above should allow for understanding of neurological mechanisms that control postural responses to expected or unexpected destabilising factors. These studies are a continuation of previous research on the analysis of motor behavior using only visual disorders [38], [39], [40]. Tests concerning correlations between selected parameters identifying responses to perturbation and postural changes as well as changes of muscular activity triggered by APAs facilitate the development of strategies to maintain both balance and readiness of quick responses to imbalance. Well-trained balance and the ability of motoric reactions to balance disorders make it possible to minimise the risk of falling and sustaining injuries in representatives of all age groups involved in daily activities as well as in relation to sports people in events characterised by particularly high susceptibility to falling [19], [38], [39]. Diagnostics, informed from the reactions of healthy individuals, will be developed for individuals experiencing problems with their lower limbs and neurological system dysfunction resulting in balance-related mechanism disorders. Widely defined prediction of falls and the further development of traditional methods supporting diagnostics and therapy constitute a utilitarian objective of this research [3], [20]. The research results might make it possible, among other things, for the development of both rehabilitation methods after injuries and fall prevention training.

2. Materials and methods

2.1. Study group

The study group included 38 participants (27 females and 11 males) aged 23 ± 2.6 years, with an average height of 172 ± 9.6 cm and an average weight of 70 ± 17 kg. None of the participants had a history of an extreme lower limb injury nor suffered from motor system dysfunctions or balance disorders.

This study was previously approved by the Ethics in Research Committee of the Academy of Physical Education in Katowice (report number 5/2020).

2.2. Experimental procedure

The measurement stand consisted of a platform enabling measurements of the distribution of pressure exerted by feet on the ground (WinFDM-S, Zebris Medical GmbH, Germany, sampling frequency of 100 Hz, 2560 tension meter sensors, sensors area = 34 × 54 cm), a treadmill for training and preventing postural perturbation (BalanceTutor, MediTouch, Israel), a cordless set for electromyography (EMG, Ultium EMG, Noraxon, USA, sampling frequency of 2000 Hz) and a cordless set of IMU (Inertial Measurement Units) sensors (Ultium Motion, Noraxon, USA, sample frequency 200 Hz). The stabilographic platform was located in the central part of the belt of the treadmill for training and preventing postural perturbation and it was attached using a two sided tape to the treadmill. In front of the platform, in a special grip, there was an IMU sensor used for the detection of each movement of the treadmill.

All systems were synchronized with each other with the use of Noraxon MR3 (Noraxon, USA) devices and the designed synchronization adapter using the mini

Fig. 1. Measurement stand with a test participant and the treadmill with the sensor and direction of perturbation

M5stack (development platform, Shenzhen, China) device with the ESP32 microcontroller. Start and stop signals for all devices are provided by the MR3 device. The developed proprietary software for M5stack allows for quick detection of the moment of treadmill movement (the device has an IMU sensor) and transmission of the signal to Noraxon software, which will ensure quick detection of the moment of treadmill movement (reaction time less than 100 μs).

The measurement stand with a patient on it is presented in Fig. 1.

2.3. Arrangement of sensors

The electromyographic tests focused on the four most important groups of muscles responsible for the maintaining balance [12], [33] – musculus tibialis anterior (TA), musculus rectus femoris (RF), musculus gastrocnemius medialis (GM) and musculus gastrocnemius lateralis (GL). Electromyographic tests of the above-named muscles are often used in the investigation of APAs and CPAs. In their research, Curuk et al. [8], [9] measured APAs using unexpected external stimuli and those initiated by a test participant. The authors demonstrated the occurrence of APAs the activity of TA, RF, GM GL. Similar conclusions were reported by Garcez et al. [11] and Krishnan et al. [21].

Fig. 2. Arrangement of EMG electrodes

The tests involved the previously mentioned muscles on both the right and left side of the body. Disposable electrodes for surface electromyography were located on the skin near bellies of the muscles, making allowances for innervation areas – guidelines for innervation zones [10], [16], [34]. The EMG electrode attachment areas were prepared by removing hair and degreasing the skin surface using ethyl alcohol. Afterwards, the electrodes were connected to cordless sensors (Fig. 2).

Additionally, the tests involved using special straps and stickers to attach 17 IMU sensors located all over the body – head, torso along the spine and shoulder, and on the upper and lower limbs. The sensors (accelerometer, magnetometer and gyroscope) were attached to the patient in accordance with the arrangement diagram provided by the producer on the webside [44]. The location of the sensors used in the research for the synchronization and determination of limb movements is shown in Fig. 3.

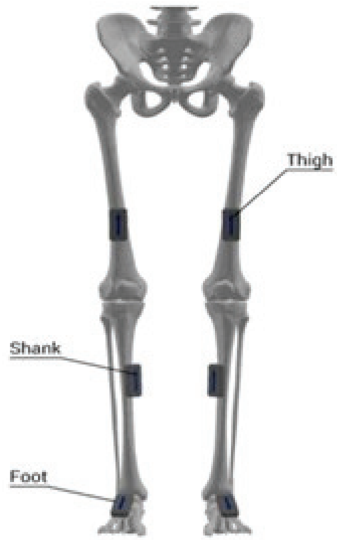

Fig. 3. Arrangement of IMU sensors

2.4. Tests

The testing procedure consisted of two stages, rest and perturbation. During the first stage (rest), the test participant was requested to sit on a chair, place their feet flat on the ground and relax their lower limb muscles (ERx). The test lasted 15 seconds and recorded the activity of the lower limb muscles at rest.

The second stage of investigation involved the use of the treadmill for training and postural perturbation prevention. The test participant was instructed to take off their footwear and enter the stabilographic platform located on the treadmill. Test participants were protected against falls by wearing a special harness attached to the treadmill. A given test participant was supposed to stand still, with their face directed forward and arms lowered freely along the sides. The tests involved three trials (Tr), i.e., Tr1, Tr2 and Tr3. Each trial consisted of two treadmill movements – forward and backward (Fig. 4). The first movement was always forward and was initiated 10 seconds after the start of the measurements. The backward movement of the treadmill was initiated 20 seconds after the start of the measurements. Both forward and backward movements of the treadmill amounted to 9.5 cm and lasted 0.52 seconds.

Breaks between the end of one perturbation and the beginning of another lasted 10 seconds. In the event of the perturbation-triggered loss of balance, the test participant was tasked with returning to the upright position as soon as possible. The test was repeated three times. During the first trial (Tr1), the participant did not know the nature, time and the direction of perturbation. During the second test (Tr2), the participant knew the nature of perturbation but not its time and direction. During the third test (Tr3), the subject knew the time left (countdown) to the occurrence of perturbation and its direction (from a dedicated display). A schematic diagram presenting the successive measurements is presented in Fig. 4.

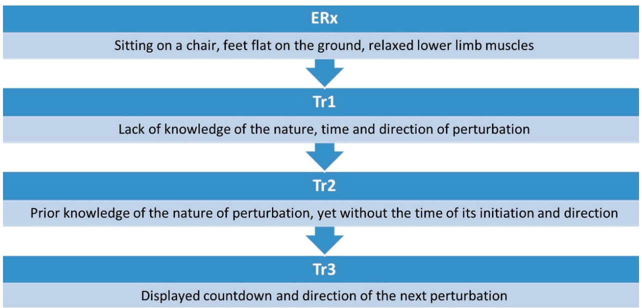

Fig. 4. Sequence of measurements

2.5. Analysis of the results

The data exported from the Noraxon software was preprocessed in accordance with the algorithm embedded in the software (MyoResearch 3.18) – the absolute values were calculated and filtered with a moving average filter with a 50 ms window.

At the first stage of the analyses, based on the results obtained in the ERx test (during sitting on a chair), it was possible to identify the average resting activity of each of the muscles subjected to analysis (EMGRx). Afterwards, the activity of each muscle measured in Tr1, Tr2 and Tr3 were divided by the EMGRx value, performing the standardisation of measured parameters in relation to the values measured at rest.

Afterwards, on the basis of the data obtained from the IMU sensor located on the treadmill belt, it was necessary to determine the beginning of the treadmill movement (t_0) for forward and backward movements.

The subsequent step aimed to answer the question whether information about the starting time and the direction of perturbation leads to an increase in the muscular tension of lower limb muscles. That stage required the investigation of the muscular activity of lower limbs in relation to APAs and EPAs in tests Tr1, Tr2 and Tr3. Time values and activity areas subjected to the analysis are presented in Fig. 5.

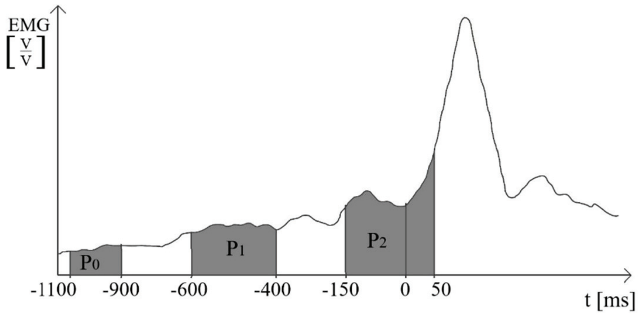

Fig. 5. Analysed time intervals of muscular activity.
The vertical axis represents the multiplicity of the resting value assuming a symbolic value of V/V.
The horizontal axis represents time in milliseconds (ms)

Area P_0 stands for the area of muscular activity during free standing (between 1100 ms and 900 ms before perturbation). Area P_1 represents the area of search for an increase in the muscular activity triggered by EPA (between 600 ms and 400 ms before perturbation). Area P_2 represents the area of an increase in the muscular activity triggered by APAs (between 150 ms before perturbation and 50 ms after perturbation). The muscular activity related to APA and EPA (EMG_{APA} and EMG_{EPA}), was determined using Eqs. (1)–(5). The values of EMG_{APA} and EMG_{EPA} were identified for each of the muscles subjected to the tests.

$$P_0 = \int_{-1100}^{-900} EMG\,dt\,, \qquad (1)$$

$$P_1 = \int_{-600}^{-400} EMG\,dt\,, \qquad (2)$$

$$P_2 = \int_{-150}^{50} EMG\,dt\,, \qquad (3)$$

$$EMG_{EPA} = P_1 - P_0\,, \qquad (4)$$

$$\mathrm{EMG}_{\mathrm{EPA}} = \mathrm{P}_2 - \mathrm{P}_0 . \quad (5)$$

Next, to determine whether the occurrence of lower limb muscle tone at the initial phase (EPA) led to an increase in the muscle tone before perturbation (APA), it was necessary to identify correlations between all of the above-named parameters in relation to all the muscles subjected to analysis in test Tr3.

After the initial observation of the increased muscular activity before perturbation, in relation to test Tr3, it was necessary to determine whether APAs did not result from the postural change connected with the forward or backward displacement of the COP and whether it did not result from the change in the flexion of the knee joint. To this end, it was necessary to calculate values of pressure exerted by the forefoot on the ground and values of the angle of flexion in the knee joint in relation to test Tr3. It was also necessary to identify the existence of correlations between the above-named values and APAs.

Finally, it was necessary to verify whether an increase in the APA-related muscular activity affected the displacement and the velocity of the COP after unbalance. The above-named displacement was always reverse in relation to the direction of the treadmill motion. The velocity of the COP was analysed as an increase in displacement 150 ms after perturbation, whereas the displacement of the COP was analysed as the maximum deviation of the COP after perturbation, regardless of its moment of occurrence, yet not later than two seconds after perturbation.

2.6. Statistical analysis

All analyses were performed using Matlab R2022a. The Shapiro–Wilk test was used to determine data normality of the parameters analysed. The lack of normal distributions was observed for Tr1, Tr2 and Tr3 tests. Based on this, it was used Friedman test followed by pairwise Wilcoxon post-hoc test with Holm correction. Additionally, the size of the effect was calculated with the methodology proposed by Rea and Parker [35]. If the difference is statistically significant ($p < 0.05$), but the effect is small, then this significance is due to reasons other than significant differences in the distribution. The difference between the medians was reported as statistically significant as long as the effect was high or at least medium. Small effects were not reported. Due to the lack of normal distributions, the Spearman correlation was calculated and presented as the correlation coefficient.

3. Results

The obtained results allowed for an objective determination of the muscle tone gain only when the examined person knew the moment of the disorder's occurrence. The research attempts to define the strategy of postural preparation and its influence on the results of postural compensation after the disorder. It was determined what could influence the speed of postural reactions and the correlation between the measured values was investigated.

3.1. Analysis of the muscular activity triggered by APAs and EPAs

Test results concerning the APA-triggered muscular activity are presented in Fig. 6 whereas those

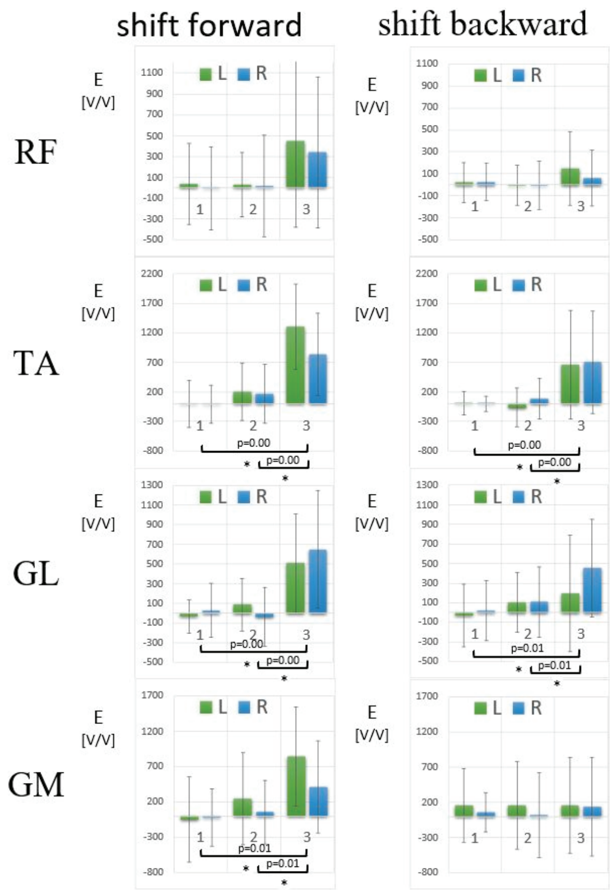

Fig. 6. EMGAPA in relation to selected RF (rectus femoris), TA (tibialis anterior), GL(gastrocnemius lateralis) and GM (gastrocnemius medialis) muscles and successive tests: Tr1, Tr2 and Tr3. Unit [V/V] indicates the multiplicity of the value of muscular activity at rest; *L* – value concerning the left lower limb and *R* – value concerning the right lower limb; Shift forward – forward movement of the treadmill belt, Shift backward – backward movement of the treadmill belt; values represent mean * standard deviation

related to the EPA-evoked muscular activity are presented in Fig. 7. Because of reference to the average value at rest, the test values were expressed in v/v. The aforesaid unit represents muscular activity constituting the multiplicity of the resting value.

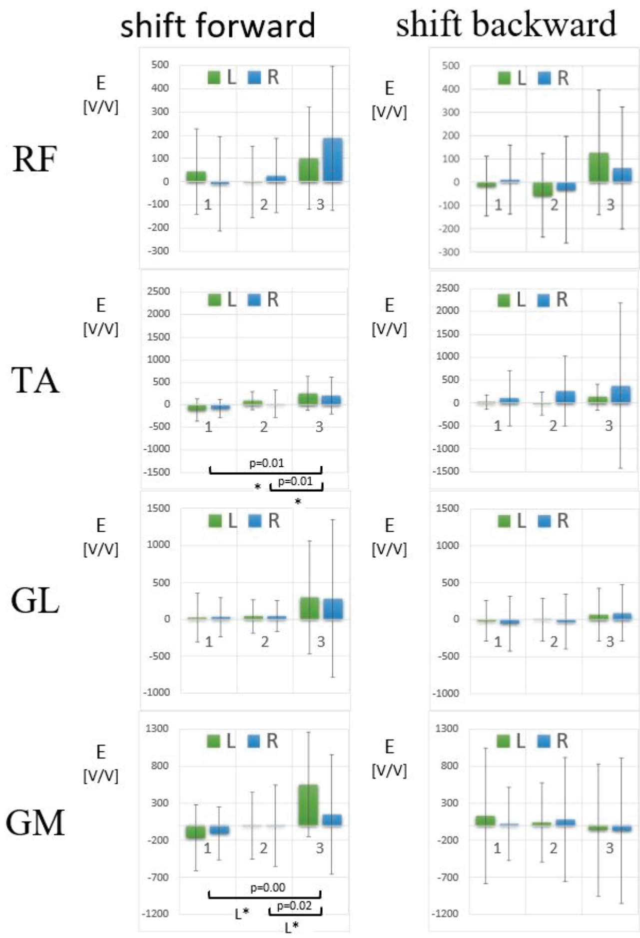

Fig. 7. EMGEPA in relation to selected RF (rectus femoris), TA (tibialis anterior), GL (gastrocnemius lateralis) and GM (gastrocnemius medialis) muscles and successive tests: Tr1, Tr2 and Tr3. The unit [v/v] indicates the multiplicity of the value of muscular activity at rest; *L* – value concerning the left lower limb, *R* – value concerning the right lower limb, Shift forward – forward movement of the treadmill belt, Shift backward – backward movement of the treadmill belt; values presented as mean * standard deviation

Statistically significant differences EMG_{APA} and EMG_{EPA} between tests Tr1 and Tr3, and tests Tr2 and Tr3 were only observed in relation to the forward shift for TA, GL and GM in relation to analyses indicating the APA phase and for TA and GM (only the left lower limb) in relation to the analyses indicating the EPA phase. The test results concerning the above-named muscles point to test Tr3 as the one revealing a significant increase in muscular activity. EMG_{APA} and EMG_{EPA} obtained in Tr3 were investigated for the correlation of muscles subjected to the tests. Correlation coefficients were determined in relation to muscles revealing varying activity between Tr3 and the remaining tests, excluding RF. The results, presented in Table 1, indicate the existence of the statistically significant correlation between the phase of the Early Postural Adjustments (EPAs) and the Anticipatory Postural Adjustment (APAs) in relation to the backward shift. The correlation could indicate that the muscle was activated earlier than the remaining muscles and that its activation increased in time until the occurrence of perturbation.

3.2. Impact of the increased muscular activity on forefoot pressure and the APA-triggered change of the angle of flexion in the knee joint

To analyse whether the increase in the muscular activity between the phase of free standing and that of the search for APAs resulted in postural changes it was necessary to investigate the correlation between the calculated value of APA and the value of increased forefoot pressure as well as the correlation between APA and the change in the value of the angle of flexion in the knee joint. Changes of forefoot pressure and of the angle of flexion in the knee joint were calculated in time intervals adopted as free standing and in the APA area (difference between the value recorded 50 ms after perturbation and the value recorded 1100 ms before perturbation). During analyses, the RF-related values were ignored as its activity was

Table 1. Results of Sperman's correlation in relation to muscular activity in the APA and EPA phase. The *p* values for the significance of the correlation are given in parentheses. The statistically significant values are bolded

	TA_L	TA_R	GL_L	GL_R	GM_L	GM_R
Shift forward	**0.73 (*p* = 0.01)**	**0.70 (*p* = 0.02)**	0.44 (*p* = 0.27)	**0.28 (*p* = 0.03)**	**0.47 (*p* = 0.02)**	0.21 (*p*=0.17)
Shift backward	**0.72 (*p* = 0.01)**	**0.69 (*p* = 0.01)**	0.33 (*p* = 0.30)	0.27 (*p*=0.35)	**0.24 (*p* = 0.04)**	**0.33 (*p* = 0.04)**

Table 2. Results of Sperman's correlation for increased muscular activity during the APA phase and increased forefoot pressure as well as for increased muscular activity during the APA phase and changes in the knee joint flexion. The *p* values for the significance of the correlation are given in parentheses. The statistically significant values are bolded

	APA – an increase in the load on the forefoot					
	TA_L	TA_R	GL_L	GL_R	GM_L	GM_R
Shift forward	**–0.28 (p = 0.01)**	**–0.33 (p = 0.01)**	0.37 (p = 0.44)	**0.27 (p = 0.01)**	0.13 (p = 0.14)	**0.43 (p = 0.01)**
Shift backward	0.22 (p = 0.22)	0.43 (p = 0.71)	0.17 (p = 0.40)	–0.27 (p = 0.71)	**0.48 (p = 0.01)**	**0.42 (p=0.01)**
	APA – flexion in the knee joint					
	TA_L	TA_R	GL_L	GL_R	GM_L	GM_R
Shift forward	0.11 (p = 0.54)	–0.23 (p = 0.78)	0.41 (p = 0.24)	**0.33 (p = 0.01)**	–0.24 (p = 0.11)	**–0.32 (p = 0.01)**
Shift backward	**0.33 (p = 0.01)**	**0.36 (p = 0.01)**	–0.25 (p = 0.51)	–0.49 (p = 0.67)	**0.36 (p = 0.01)**	**0.39 (p = 0.01)**

not detected in forward and backward translation. The test results are presented in Table 2.

The tests were also concerned with differences between the average values of forefoot pressure in subsequent phases of Tr1, Tr2 and Tr3. The results did not reveal any statistically significant differences between the average values. Similarly, no statistically significant differences were observed between the average values of the angle of flexion in the knee joint in successive tests: Tr1, Tr2 and Tr3.

3.3. Impact of the increased muscular activity before perturbation on the velocity and displacement of the COP after perturbation

During the APA search phase (–150 ms–(+50 ms)), a correlation between the calculated value of APAs and the velocity of the COP at the initial stage of motion as well as between APAs and the maximum displacement of the COP triggered by perturbation was observed (Table 3). The results concerning the correlation of velocity and the maximum displacement of TA revealed a strong correlation between the increased activity of the muscle and the COP displacement velocity and the COP displacement value obtained in relation to the forward motion of the treadmill (Table 3). A slightly lower correlation was obtained in relation to GL and the backward shift of the treadmill.

4. Discussion

The above-presented tests aimed to identify the impact of the forward and backward shift of the ground on the muscular response in lower limbs connected with postural adjustment. Understanding how

Table 3. Results of Sperman's correlation for increased muscular activity during the APA phase and the COP displacement velocity after perturbation as well as for increased muscular activity during the APA phase and the maximum COP displacement. The *p* values for the significance of the correlation are given in parentheses. The statistically significant values are bolded

	APA – V					
	TA_L	TA_R	GL_L	GL_R	GM_L	GM_R
Shift forward	**0.71 (p = 0.00)**	**0.73 (p = 0.00)**	0.21 (p = 0.17)	0.43 (p = 0.09)	0.51 (p = 0.34)	0.11 (p = 0.27)
Shift backward	0.62 (p = 0.51)	0.53 (p = 0.27)	**0.59 (p = 0.00)**	**0.58 (p = 0.01)**	0.12 (p = 0.37)	0.11 (p = 0.33)
	APA – COP displacement					
	TA_L	TA_R	GL_L	GL_R	GM_L	GM_R
Shift forward	**0.71 (p = 0.00)**	**0.63 (p = 0.00)**	0.21 (p = 0.13)	0.43 (p = 0.22)	0.61 (p = 0.55)	0.12 (p = 0.87)
Shift backward	0.62 (p = 0.59)	0.63 (p = 0.21)	**0.59 (p = 0.00)**	**0.54 (p = 0.00)**	0.17 (p = 0.40)	0.09 (p = 0.71)

adjustment to perturbation and the investigation of the effect of the aforesaid manner on the COP displacement after perturbation enabled the determination of the components of a strategy aimed to maintain balance. The identification of the strategy of adjustment to perturbation was restricted to the assessment of the activity of selected muscles, the evaluation of flexion in the knee joint and the assessment of pressure exerted on the ground, which was tantamount to the displacement of the COP. As reported elsewhere [1], [26], [29], [32], [36], muscular activity was used to identify Early Postural Adjustments (EPAs) and Anticipatory Postural Adjustments (APAs).

4.1. Detection of muscular activity before perturbation

The analysis of selected muscles revealed that during APA (Fig. 6) in relation to Tr3, i.e., when test participants were informed about the starting time of perturbation, it was possible to notice an increase in average muscular activity. The aforesaid increase occurred in the case of all tested muscles, i.e., RF, TA, GL and GM in the APA phase. The increase was noticeable in relation to the adjustment to both forward and backward perturbation. However, it should be noted that an increase in the average value was accompanied by an increase of the standard deviation. As a result, some increases did not reveal statistically significant differences. Statistically significant increases were observed between Tr1 and Tr3 as well as Tr2 and Tr3 in relation to the TA, GL and GM muscles for APA before the forward shift and in relation to the TA and GL muscles for APA before the backward shift. Therefore, similarly to research by Cleworthet at al. [7], it was demonstrated that the activity of TA played an important role in APA. In accordance with publications [7], [26], the above-named muscle played an important role in the perturbation-related adjustment. Scariot et al. [29] indicated TA as the muscle which was activated when adjusting to any movements connected with jumps, taking a step or catching a flying ball. Apart from making movements in the ankle joints, the TA also protects against the occurrence of flat feet [5], the occurrence of which may disturb postural reactions. This finding was confirmed by the Shein et al.'s is study, which found a reduced TA cross-section in ultrasonography in a group of patients with flat feet [42]. The greatest changes in the absolute values of muscle forces during performance of tasks related to maintaining posture under various conditions were also observed for the TA muscle [12]. The above fact would mean weaker activity of this muscle with flat feet. The presence of flat feet affects the way the body is balanced and the displacement of the COP [17]. We should also remember that, when positioned medially or laterally in the medial compartment of the extensor retinaculum, it is able to perform supination and pronation movements in the lower ankle joint and participate in postural reactions in the frontal plane by stabilizing the lower ankle joint [5], [24].

Similar conclusions related to forward shifts of the treadmill belt resulted from the analysis of GL and GM. Oeffinger et al. [26] indicated the GL and GM as muscles, whose activation was particularly important in the stabilisation of the body in a response to the temporary dysfunction of lower limbs in patients after anterior cruciate ligament reconstruction.

By observing the anatomy and topography of GM and GL, their relationship with capsular-ligament structures can be noticed in knee joint [22], [27], [28], [37]. The GM connects to the posterior medial capsule of the knee joint [22] while The GL connects to the fabellofibular ligament [27], [28], [32]. The above combination may affect the mechanosensive function of the capsule and ligaments of the knee joint and thus pre-prepare the receptors and thus the whole organism for disorders [18].

We replicated findings of Scariot et al. [29] and Lee et al. [23] indicating that activation of the TA connected with adjustment to perturbation occurred significantly earlier than in the APA phase and included EPAs. The above-presented observations were confirmed by the analysis of muscular activity in the EPA phase (Fig. 7), revealing the existence of statistically significant differences between Tr1, Tr2 and Tr3 in relation to TA and its adjustment for the forward movement of the treadmill belt. The subsequent stage involved investigation focused on the existence of correlation between increased muscular activity in the EPA and APA phases. The results presented in Table 1 indicate a linear correlation between the activity of TA obtained for both left and right lower limbs of the test participants. The foregoing revealed that the activity of the aforesaid muscles already grew during the EPA phase and the APA phase constituted the continuation of the above-named increase. The results presented in Table 1 did not reveal an increase in GL and GM activity during the EPA phase, which indicated that the muscles were activated shortly before the perturbation. Their activity increased significantly and more intensely than that of the TA and followed the EPA phase. Muscular activation could result in the movements of limbs or changes in the position of limbs, including, e.g., flexion in the knee joint or the forward inclination of the body. Perhaps this is because of the fact that the TA is a highly motorized muscle and does not connect with the ligamentous and capsular structures of the knee joint and ankle joints [5] while GM and GM connect with the ligaments and capsule of the knee joint, as mentioned earlier [22], [27], [28], [37].

4.2. Postural changes triggered by APA

Research by Cleworth et al. [7] indicated slight lower limb flexion by an angle of 0.5° during the phase of adjustment to similar perturbation connected with the forward movement of the ground. In turn, the researchers reported a lack of statistically significant differences connected with the flexion of the upper leg and trunk. The tests involving the analysis of the flexion in the knee joint did not reveal statistically significant changes between phases Tr1, Tr2 and Tr3. Similarly, no statistically significant changes were observed in the displacement of the COP nor in forefoot pressure. Therefore, it could be supposed that a postural adjustment to perturbation was not connected with forward or backward trunk inclination and that the observed increase in the tension of TA, GL and GM did not affect the flexion in the knee joint. The foregoing conclusion was confirmed by the results presented in Table 2, indicating the lacking correlation between muscle tone and the flexion in the knee joint and the increase in forefoot pressure. The increased muscle tone combined with the lack of the COP displacement may indicate the block of the joint, which could translate into a longer adjustment to perturbation or indicate other postural changes, such as backward pelvic girdle inclination and forward shoulder girdle inclination, which requires tests in the scope of kinematics of the upper part of the body.

4.3. Impact of APA on the selected muscle reactions to postural compensation after perturbation

The analysis of correlations between muscle tone and the COP displacement as a well the velocity of the COP after perturbation (Table 3) indicated a strong correlation between the muscle tone of the TA and the COP displacement as well as the displacement of the COP after perturbation. In terms of the GL, the abovenamed correlation was lower. In both cases, the observed increased muscle tone resulted in the extension of the path and velocity of the COP displacement. The aforesaid results, confirmed observations by Mohapatra et al. [25], report tests involving hitting a body trunk with a weight led to the increased displacement of the COP in measurements when the test participant was informed on the occurrence of perturbation. The results could indicate that the increased muscle tone in the APA phase was responsible for block in the joints before perturbation, which, in turn, changed the pattern of postural compensation after perturbation. The higher COP moved significantly further than in the situation when muscle tone did not occur and increased both path and velocity-related values.

5. Conclusions

The data explicitly indicated that the factor triggering postural adjustment and readiness for response to perturbation was the knowledge of the expected time of the perturbation. These data did not reveal differences in the activation of selected muscles, differences in flexion in the knee joint as well as differences in changes of backward-forward pressure between the tests where the participants did not know or knew the nature of perturbation but were unaware of its starting time.

An important factor connected with the adjustment to perturbation was muscle tone responsible for blocking of joints of lower limbs and, consequently, a greater inclination of the COP after perturbation.

The subsequent stages of tests should focus on the search for objective methods enabling the measurements of body kinematics resulting from postural compensation as a response to perturbation and their relationship with the adjustment to perturbation. It is also important to overcome some limitations that can improve the accuracy of measurements. Such limitations include the inability to collect tissues and determine the number of receptors in the muscles, the lack of a needle method for EMG measurement and the lack of isokinetic studies performed. These findings may generate training programme of muscles and body stability aimed to reduce the incidence of traumas and falls due to unexpected perturbation of the body and ground. The determination of a strategy enabling the maintaining of balance before perturbation based on muscular activity and values of ground reaction forces could provide information about proper patterns of response to perturbation. These data may also provide guidelines for persons suffering from balance disorders due to diseases and injuries.

Conflict of interest

The authors declare no conflict of interest. The authors declare that all authors were fully involved in the study and the preparation of the manuscript and that the material within has not been and is not to be submitted for publication elsewhere.

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The stock market indexes in research on human balance

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Purpose: The aim of the study was to demonstrate the possibility of using stock exchange indices to assess the ability to maintain balance as a supplement to analyzes using values determined in the time and frequency domains. *Methods*: 83 healthy people (56 females, 27 males, age years 21 SD = 1.3 years) participated in the research. Measurements were performed with open and closed eyes and in the virtual environment with two sceneries oscillating at four frequencies. The results determined in the time and frequency domains were analyzed in relation to the results calculated with the use of stock exchange indicators for which the Trend Change Index was formulated. Performed measurements made it possible to determine the average COP speed, the average COP speed and range of movement towards AP, power spectral density PSD and stock exchange indices. *Results*: In the case of PSD values for the ranges above and below 0.5 Hz, statistically significant differences occurred for most measurements. Obtained values of TCI coefficient were similar and no statistically significant differences were observed. The maximum values of the PSD medians were obtained in trials with the oscillating scenery. *Conclusions*: Conducted analyzes showed that use of stock exchange indicators broadens the interpretative possibilities of COP measurements by determining the number of consecutive skips (changes in the direction) of the COP and prioritizing according to the times between them. The applied stock market analysis methods also filtered out changes in the position resulting from noises that could not be removed with the use of standard low-pass filters.

Key words: *postural stability, virtual reality, center of pressure, stock market indicators*

1. Introduction

Keeping the body in a standing position requires constant control and correction of the position, velocity and acceleration of the body's center of mass (COM) [9], [10]. Searching for the information about the strategies and mechanisms describing the process of maintaining balance has resulted in numerous approaches to analyzing center of pressure of foot on the ground (COP) changes – a value which is easy to measure and interpret. The most frequently used methods define these changes in relation to time [1], [6], [24], e.g., path length, the average COP velocity ellipse area or ranges in anteroposterior (AP) and mediolateral (ML) directions. As standard, it is assumed that the higher the value of the aforementioned quantities, the worse the ability to maintain balance [3], [4], [11]. Since some repetitive sequences can be seen in the COP displacement signal course, frequency domain analyzes are a natural extension of time domain analyzes. They make it possible to determine which cyclic components the analyzed signal consists of, thus complementing analyzes in the time domain [7], [13], [23]. Research in the virtual environment can be mentioned as an example of measurements in sensory conflict conditions where the oscillating virtual scenery was used with the still, real floor. Such cyclic movement of the surroundings causes changes in the body movement pattern, what changes COP motion. Then, for example, the increase in path length of COP can be explained by means of frequency

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analysis – one can check what cyclic components are part of the movement and whether the following scenery is a dominant one [12], [21]. Other methods, such as wavelet analysis [8], [17], [22] or entropy determination [5], [16], are used much less often.

However, frequency domain analysis is limited to changes that are cyclical in nature and only such changes in COP positions are detected. However, changes that are not of this nature are ignored. The ability to detect these types of changes – not cyclical – seems to be particularly important in terms of assessing the balance maintaining strategy. As proven, there is a relation between the frequency of COP movements and the strategy of maintaining balance, where low frequencies (0–0.5 Hz) mostly account for visuo-vestibular regulation, medium frequencies (0.5–2 Hz) for cerebellar regulation and high frequencies (>2 Hz) for proprioceptive regulation [2], [15]. However, speaking of frequency, it should be remembered that, in the case of successive COP positions, it should be interpreted as the rate of these changes, i.e., the time between individual changes in the COP position [20], [25]. With such an interpretation, it can be seen that determining the actual number of consecutive changes in the position of the COP, and not only those of a cyclical nature, becomes crucial [14], [19]. Moreover, the applied method should eliminate changes in the position that are too small – of a short time, which may constitute noises, and which remain in the signal even after the application of low-pass filters.

The aim of the study was to demonstrate the possibility of using stock exchange indices to assess the ability to maintain balance as a supplement to analyzes with the use of values determined in the time and frequency domains. The use of stock exchange indicators is to allow the detection of significant changes in the trend – changes in direction – in the COP movement along with the determination of the duration between consecutive jumps. Such an analysis should enable us, for a certain frequency range, to detect both cyclical components and non-cyclical changes

2. Materials and methods

Participants

The participants consisted of 83 healthy people (56 females and 27 males) with an average age of 21 years (SD = 1.3 years), mean body mass of 65 kg (SD = 12.2 kg) and an average height of 172 cm (SD = 8.3 cm). Health problems relating to maintaining balance and obesity (body mass index; BMI > 30) were considered as exclusion criteria. This study was first approved by the Ethics in Research Committee of the Academy of Physical Education in Katowice (protocol number 5/2020). In accordance with the Ethics in Research Committee, each participant provided informed consent regarding their participation in the study. Consent to participate in the study was expressed verbally (written consent was not required).

Measurement stand

Measuring equipment consisted of a platform for measuring the distribution of pressure exerted by the foot on the ground (WinFDM-S, Zebris, sample frequency of 100 Hz, 2560 tensiometer sensors, sensing area = 34 cm × 54 cm) and an Oculus Rift Head Mounted Display (HMD) for projecting spatial images. The virtual sceneries, which were projected during testing, were prepared in the Unity 3D environment. Two sceneries were prepared: the first scenery was termed "closed" (Fig. 1a) and consisted of a furnished room in which objects and the wall were seen by the participant at a distance of approximately 3 m; the second scenery was termed "open" (Fig. 1b) and consisted of a desert scene with objects seen at a distance of approximately 100 m.

Fig. 1a. View of closed scenery

Fig. 1b. View of open scenery

During measurements, the sceneries oscillated in the sagittal plane at constant frequencies. Participants saw the movement as movement of the entirety of the surroundings. The movement of the scenery was achieved by overlapping the following two movements:

- movement along the sagittal axis (AP direction) of the currently tested participant (Eq. (1))

$$y = A_y * \sin(2\pi f_t t), \qquad (1)$$

where:

f_t – frequency of the scenery movements,

A_y – amplitude of the scenery movements in centimeters,

t – time;

- movement around the transverse axis of a given test participant (Eq. (2))

$$\phi = A_\phi * \sin(2\pi f_r t), \qquad (2)$$

where:

f_r – frequency of the scenery movements,

A_φ – amplitude of the scenery movements in degrees,

t – time.

The following designations were adopted for the purpose of the tests utilized:

- $A_y = 15$ [cm],
- $A_\varphi = 1$ [deg],
- $f_r = 0.5 * f_t$,
- f_t (depending on test condition) = 0.2 Hz or 0.5 Hz or 0.7 Hz or 1.4 Hz.

Each test lasted 30 s. There was one trial for each condition of the measurement.

Experimental procedure

The procedure consisted of tests performed in the real and virtual environments. In the real environment, measurements were performed with eyes open (EO) and eyes closed (EC). Measurements in the virtual environment were performed using two sceneries ("open" and "closed") that oscillated with constant frequencies of 0.2 Hz, 0.5 Hz, 0.7 Hz and 1.4 Hz. Participants were required to take off their shoes and step on the measuring platform in an upright position. During measurements participants had to stand still with their arms crossed over their chests and their eyes focused on a designated point. Measurements were performed in the real environment followed by measurements in the virtual environment (Fig. 2). Participants were divided into four groups:

- "open" scenery at frequencies of 0.2 Hz (0.2 O) and 0.5 Hz (0.5 O),
- "open" scenery at frequencies of 0.7 Hz (0.7 O) and 1.4 Hz (1.4 O),
- "closed" scenery at frequencies of 0.2 Hz (0.2 C) and 0.5 Hz (0.5 C),
- "closed" scenery at frequencies of 0.7 Hz (0.7 C) and 1.4 Hz (1.4 C).

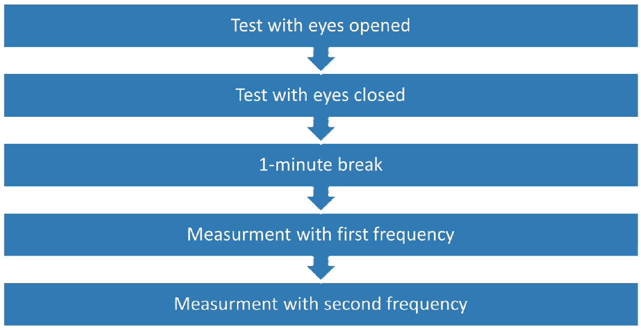


Fig. 2. Sequence of one measuring series

Each participant was tested in the real environment and then in a virtual reality environment:

- for a group of 40 participants, measurements were obtained in "open" scenery; 24 participants from this group took part in tests with scenery movement frequencies of 0.2 Hz and 0.5 Hz, with the remaining 16 participating in tests with scenery movement frequencies of 0.7 Hz and 1.4 Hz.
- for a group of 43 measurements were obtained in "closed" scenery, with 32 participants tested at scenery movement frequencies of 0.2 Hz and 0.5 Hz and 11 participants tested at scenery movement frequencies of 0.7 Hz and 1.4 Hz

Analyzed quantities

The measurements obtained were the displacements of the COP for successive moments during the 30 s of testing. Analysis was performed using MATLAB software. In the first stage, analysis of the average velocity of the COP – total and in AP direction – and the COP range of motion in AP direction was performed. The results were compared for all conditions. In the next step, the frequency analysis of the measurements was performed by determining the PSD of the COP displacement in the AP direction, determined as the square of the amplitude values of the bands from the (FFT) calculations divided by the unit frequency. Then, the Trend Change Index (TCI) coefficient was calculated. This coefficient defines the number of changes in the trend, determined as the number of signal intersections from the Moving Average Convergence Divergence (MACD) calculation algorithm.

Calculation of TCI and MACD coefficients

The MACD is presented in the form of two lines: the MACD line and the signal line. The MACD line was obtained by subtracting two moving averages with exponential weights taking into account 12 samples (short-period) and 26 samples (long-period) (Eqs. (3) and (4)).

$$\mathrm{MACD}_{12,26} = \mathrm{EMA}_{12} - \mathrm{EMA}_{26}, \tag{3}$$

where:
EMA_{12} – faster exponential moving average,
EMA_{26} – slower exponential moving average;

$$\mathrm{EMA}_{pN} = \frac{p_0 + (1-\alpha)p_1 + (1-\alpha)^2 p_2 + (1-\alpha)^3 p_3 + \ldots + (1-a)^N p_N}{1 + (1-\alpha) + (1-\alpha)^2 + (1-\alpha)^3 + \ldots + (1-\alpha)^N}, \tag{4}$$

where:
p_0 – ultimate value,
p_1 – penultimate value,
p_N – value preceding N periods,
N – number of periods.

The signal line was obtained by calculating the moving average with exponential weight for the MACD signal taking into account the nine MACD signal samples (Eq. (5)).

$$\text{Signal Line} = \mathrm{EMA}_{\mathrm{MACD},9}. \tag{5}$$

The intersection of the MACD and signal line signals marks a trend change in the COP displacement signal in the AP direction. A graphical interpretation of the crossings for the COP signal is shown in Fig. 3. Changes in the trend are related to the change in the direction of the COP movement. The TCI coefficient is the number of changes in the signal trend during a 30 s test, calculated as the sum of the MACD and signal line crossings.

Comparative analysis of the PSD and MACD

The calculated values of the PSD and MACD coefficients were used to analyze the influence of the disturbances participants were exposed to. The analysis compared both the effects of particular scenery and the frequency of oscillations on the participants, as well as the differences in relation to the real environment. Both PSD and MACD calculations were made for the AP direction.

The MACD coefficient made it possible to determine the number of changes in the trend throughout the entirety of every 30 s measurement. Detailed analysis showed that timespans differed between individual changes in the trend. Therefore, the detected trend changes were grouped according to the time preceding these changes and according to the ranges indicated in row T of Table 1. To make comparative analysis of the MACD ratio with the calculated PSD possible, the time intervals obtained for MACD were converted into corresponding frequencies using the following assumptions:

- the time of the COP skips between two consecutive positions detected as a trend change was treated as a half of the whole cyclic movement in the forward or backward direction;
- in the case of a cyclical movement, the time before the detected trend change was half of the full cycle (i.e., it takes half of the period); hence, the period was twice the measured time preceding the detected trend change;

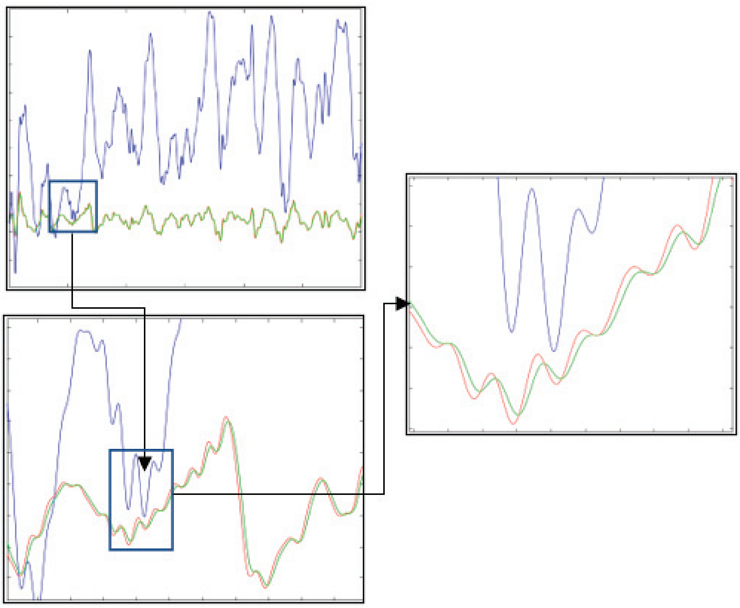


Fig. 3. Signal line and MACD waveforms for an exemplary COP waveform.
The intersections of the MACD and signal line signals mark the locations of the trend changes in the COP signal

Table 1. The time intervals considered during the analyzes
and the corresponding frequency intervals of the COP displacement signal

T [s]	0.05–0.1	0.1–0.2	0.2–0.3	0.3–0.4	0.4–0.5	0.5–0.6
f [Hz]	5.0–10.0	2.5–5.0	1.67–2.5	1.25–1.67	1.0–1.25	0.83–1.0
T [s]	0.6 – 0.7	0.7–0.8	0.8–0.9	0.9–1.0	0.05–1.0	1.0–30.0
f [Hz]	0.71–1.0	0.625–1	0.56–0.625	0.5–0.56	>0.5	<0.5

- the frequency of such a cyclic component was calculated in accordance with Eq. (6).

$$f = \frac{1}{2t_p}, \tag{6}$$

where:

f – frequency,

t_p = time between consecutive detected trend changes with $2t_p$ = period.

It should be emphasized that detected changes in the trend can be both cyclical and non-cyclical, whilst the presented conversion is intended to group the obtained results of the calculations in a manner that enables the comparative analysis of MACD and PSD. The summary of the analyzed time intervals and the corresponding frequencies is presented in Table 1. During analyzes of PSD values the total power was determined for ranges of 0.1–0.5 Hz and 0.5–10 Hz, as well as for 0.5–10 Hz divided into intervals, as shown in Table 1. MACD coefficients were determined in total for the range 0.5–10 Hz and with division into intervals according to Table 1. The MACD coefficients for the range below 0.5 Hz for the parameters adopted in the calculations were zero. This means that no trend changes where the time was longer than 1 s were detected.

Statistical analysis

The statistical analysis of the results was performed with use of the Statistica 13 software. In the first step of the analysis, the occurrence of normal distributions for each of the compared values was tested for using the Shapiro–Wilk test. The results did not confirm the presence of normal distributions in all subgroups, hence, the graphs show the median for the calculated values. In the next step, the Kruskal–Wallis ANOVA test and Dunn's post-hoc test were performed to investigate whether there were significant statistical differences in the analyzed groups.

3. Results

The measured values of successive COP positions in time were developed using analyzes in time and frequency domains and on the basis of algorithms for determining stock exchange coefficients. In Table 2, the basic time-domain calculations that are among the most frequently analyzed variables in balance maintenance ability tests (i.e., the resultant average COP velocity, the average COP velocity in the AP direction and the range of motion in the AP direction) are shown. The results presented from analyzes in the time domain show a statistically significant increase in the values measured during the test in the virtual environment in relation to tests in the real environment for all measurements except for:

- results of the average total COP velocity and average velocity in the AP direction when comparing the measurements with EC with measurements in "open" and "closed" scenery with oscillation frequencies of 0.2 Hz;
- the range of motion in the AP direction when comparing measurements made in the real environment (both EO and EC) with measurements in "open" scenery with oscillation frequencies of 0.2 Hz and 0.5 Hz.

In Figure 4a, the calculated PSD values for the ranges above and below 0.5 Hz are shown. Despite the noticeable differences between the values obtained for individual types of measurements, In Table 3, it is shown that statistically significant differences occur only in the case of:

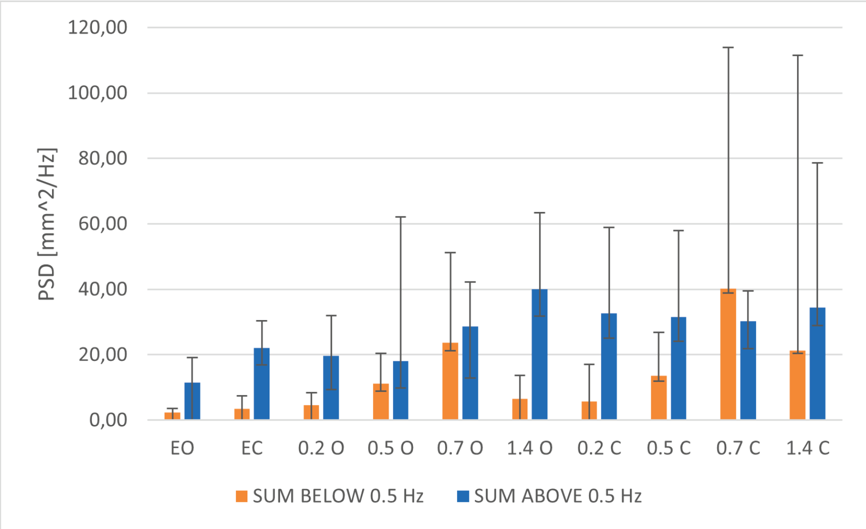
- comparisons of tests with EO in relation to all other measurements for both frequency ranges;
- the frequency range of 0.5–10 Hz, the comparison of tests with EC in relation to measurements in "open" scenery with a scenery oscillation frequency of 0.7 Hz and in "closed" scenery with scenery oscillation frequencies of 0.5 Hz, 0.7 Hz and 1.4 Hz;
- the frequency range below 0.5 Hz, comparisons of tests with EC for all measurements except for measurements in "open" scenery where the scenery oscillated at frequencies of 0.2 Hz and 0.5 Hz.

Comparison of measurements made in the virtual environment with each other did not show statistically significant differences.

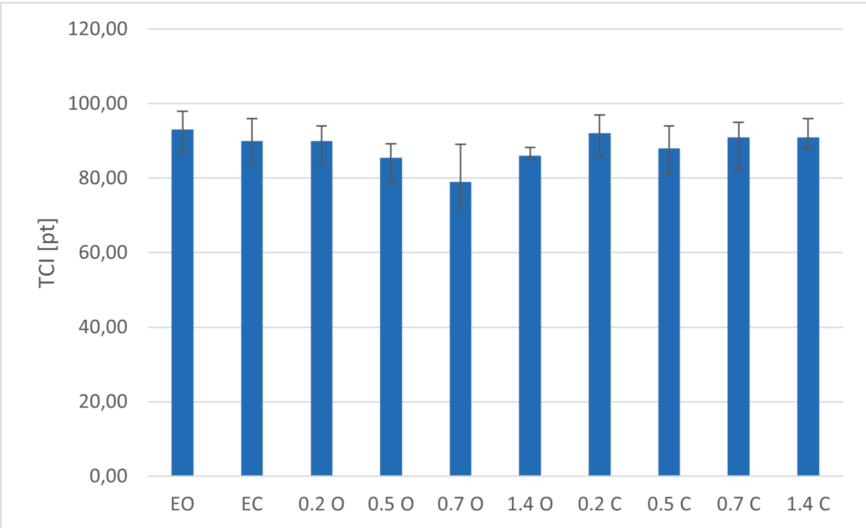
Comparing the PSD values obtained in individual intervals above and below 0.5 Hz (Table 3) showed statistically significant differences for all measurements except for measurements in "open" scenery with the

Table 2. Values obtained in the time domain: V = average velocity of Center of Pressure,
V_AP = average velocity of COP in AP direction, R_AP = range of motion of COP in AP direction

	EO	EC	0.2 O	0.5 O	0.7 O	1.4 O	0.2 C	0.5 C	0.7 C	1.4 C
V	7.6	9.0	10.8	12.0	14.2	11.6	9.7	11.4	13.7	20.5
V_AP	4.6	5.8	7.0	7.6	11.3	8.2	6.9	8.5	10.7	16.7
R_AP	17.0	17.2	19.6	19.0	27.3	23.9	21.7	23.3	38.0	31.5



(a)



(b)

Fig. 4. (a) PSD values obtained for frequencies above and below 0.5 Hz,
(b) TCI values for time intervals corresponding to frequencies above 0.5 Hz

Table 3. The p values ($\alpha = 0.05$) obtained for the comparison of the PSD results for tests in real and virtual environments

Measurement	EO	EC	0.2 O	0.5 O	0.7 O	1.4 O	0.2 C	0.5 C	0.7 C	1.4 C
	above 0.5 Hz									
EO		0.044	0.015	0.000	0.000	0.006	0.000	0.000	0.000	0.000
EC	x	x	0.926	0.201	0.001	0.613	0.418	0.000	0.000	0.000
	below 0.5 Hz									
EO		0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000
EC	x	x	0.130	0.081	0.008	0.001	0.003	0.000	0.015	0.035
	comparison between results above 0.5 Hz and below 0.5 Hz									
	0.000	0.000	0.000	0.000	0.501	0.004	0.000	0.003	0.071	0.136

Legend: EO – eyes open; EC – Eyes closed.

scenery oscillation at 0.7 Hz and in "closed" scenery with the scenery oscillation at 0.7 Hz and 1.4 Hz.

In the case of summary analyzes of the values of the TCI coefficient calculated for individual types of tests (ranges above 0.5 Hz), the obtained values are similar (Fig. 4b) and no statistically significant differences were observed.

It is noteworthy that the very large values of the coefficients of variation (CV) (Table 4) obtained in the case of PSD analyzes in relation to similar analyzes made with the use of the TCI coefficient.

Analyzing the effect size for Dunn's post-hoc test, all achieved values of the *r* (epsilon square) parameter exceed the value of 0.38. In the interpretation proposed by Rea and Parker [18] it means "strong" effect size. The result means that we can consider the results of the comparisons as valid for the current research group.

In Figures 5 and 6, the median number of trend changes recorded in individual measurements, divided into frequency ranges corresponding to the ranges in Table 1, are shown. Comparison of the measurements carried out with EO with those carried out with EC showed no statistically significant differences for any of the intervals (Table 5). When comparing measurements obtained in the real environment (EO and EC) with measurements obtained in the virtual environment, statistically significant differences were observed primarily with scenery oscillations at frequencies of 1.4 Hz and of 0.7 Hz and regardless of the type of scenery used (Table 5). When comparing the measurements made in the virtual environment with each other, statistically significant differences were found only when comparing tests performed with a scenery oscillation frequency of 1.4 Hz with tests performed at other frequencies.

In the case of the figures in which the values of median PSD from individual measurements with division into frequency ranges are shown (Figs. 7 and 8), it can be seen that the maximum values were obtained

Table 4. Coefficient of variation obtained for total PSD for frequency ranges below and above 0.5 Hz and TCI for frequencies above 0.5 Hz

CV of:	EO	EC	0.2 O	0.5 O	0.7 O	1.4 O	0.2 C	0.5 C	0.7 C	1.4 C
PSD 0–0.5 Hz [%]	57.1	37.6	49.9	142.5	38.1	49.0	53.2	58.4	24.1	91.2
PSD >0.5 Hz [%]	50.4	73.9	57.7	79.9	95.4	72.7	119.9	73.9	122.7	227.0
TCI [%]	5.9	6.7	5.5	6.1	10.7	1.9	6.0	7.4	6.9	4.4

Legend: EO – eyes open; EC – Eyes closed.

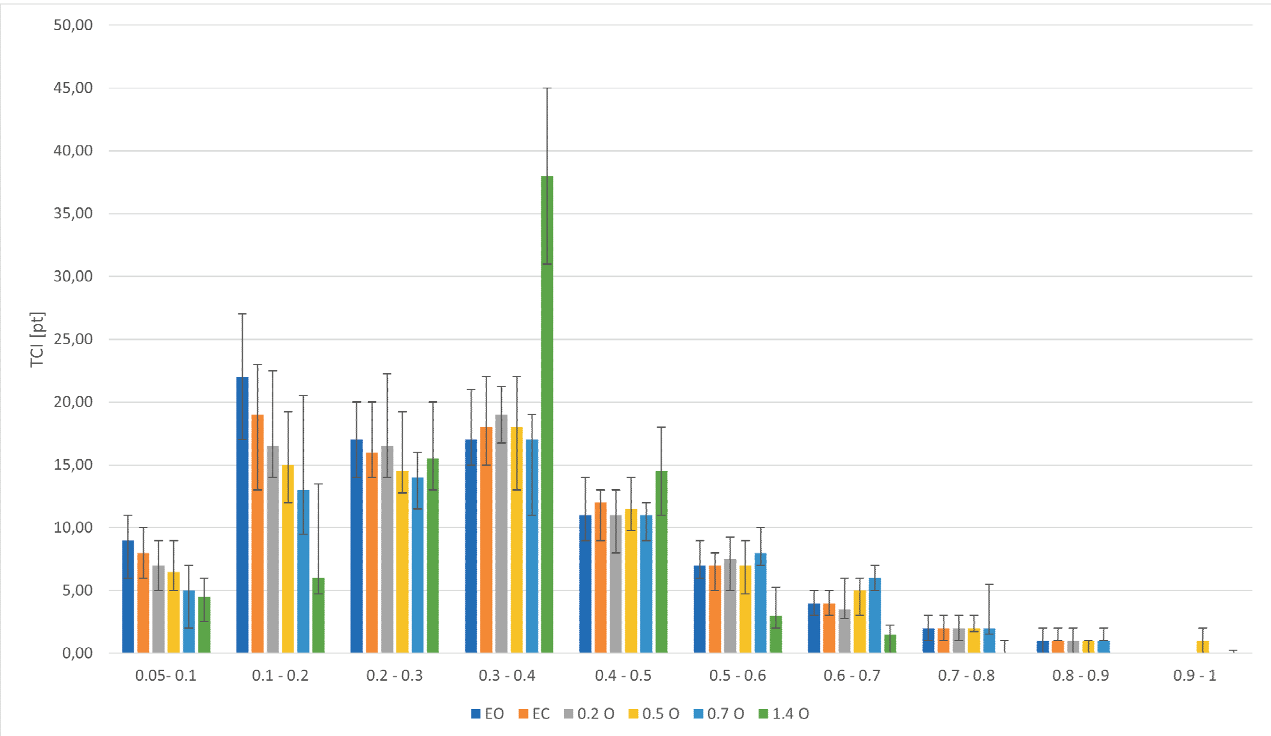


Fig. 5. TCI for individual time slots for measurements in the real environment and the open scenery

in measurements with scenery oscillation frequencies of 0.7 and 1.4 Hz in intervals corresponding to these frequencies and regardless of the type of scenery. Also, in the case of tests with the scenery oscillating at a frequency of 0.5 Hz, the increase in the PSD value in the range corresponding to this frequency is noticeable. The remaining PSD values, especially in the case of measurements in the real environment (Fig. 9), are negligibly small. The *p* values indicate statistically significant differences between the measurements in the real and virtual environment are presented in Table 6. In the case of comparing the measurements in the virtual environment between themselves, statistically significant differences were obtained only when the measurements

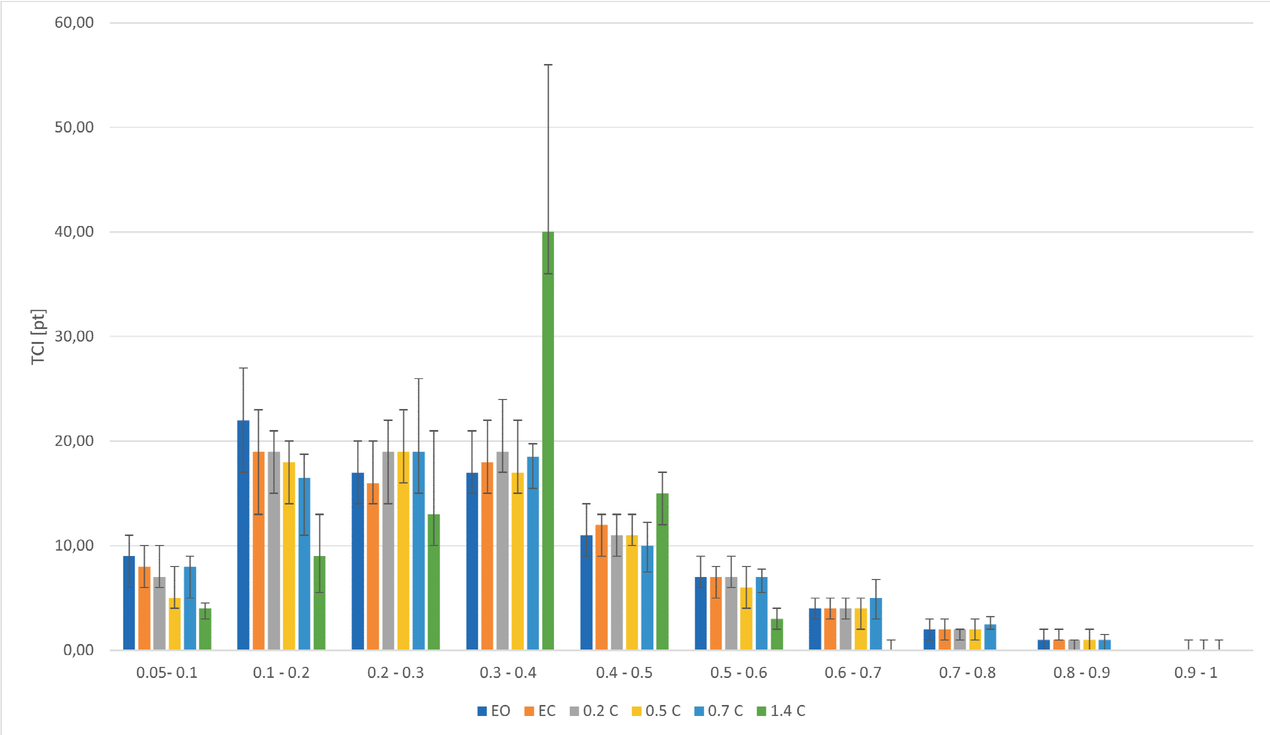


Fig. 6. TCI for individual time intervals for measurements in the real environment and in the closed scenery

Table 5. *p* values ($\alpha = 0.05$) for comparison of TCI between individual studies

Research	Time interval [s]	0.05–0.1	0.1–0.2	0.2–0.3	0.3–0.4	0.4–0.5	0.5–0.6	0.6–0.7	0.7–0.8	0.8–0.9	0.9–1.0
EO	EC	0.959	0.176	1.000	0.839	1.000	0.901	1.000	1.000	0.889	1.000
EO	0.2 O	0.484	0.308	1.000	0.861	0.994	1.000	1.000	1.000	0.985	1.000
EO	0.5 O	0.557	0.005	0.932	1.000	1.000	1.000	0.933	0.995	1.000	0.672
EO	0.7 O	0.002	0.004	0.426	0.998	0.983	0.987	0.020	0.931	0.579	0.979
EO	1.4 O	0.000	0.000	0.999	0.000	0.118	0.000	0.017	0.004	0.043	0.905
EO	0.2 C	0.808	0.370	1.000	0.493	1.000	1.000	1.000	0.978	0.999	1.000
EO	0.5 C	0.004	0.126	0.829	0.988	1.000	0.414	1.000	0.985	1.000	1.000
EO	0.7 C	0.947	0.173	0.986	1.000	0.950	1.000	0.950	0.965	1.000	1.000
EO	1.4 C	0.000	0.000	0.919	0.000	0.580	0.000	0.000	0.000	0.044	0.186
EC	0.2 O	0.951	1.000	1.000	1.000	1.000	0.979	0.999	1.000	1.000	1.000
EC	0.5 O	0.971	0.563	0.942	1.000	1.000	1.000	0.992	0.996	0.863	0.799
EC	0.7 O	0.025	0.341	0.447	0.755	0.997	0.665	0.055	0.933	0.976	0.948
EC	1.4 O	0.001	0.000	0.999	0.000	0.062	0.005	0.006	0.004	0.002	0.830
EC	0.2 C	0.999	1.000	1.000	0.995	1.000	0.823	1.000	0.978	0.684	1.000
EC	0.5 C	0.094	0.998	0.807	1.000	1.000	0.979	1.000	0.984	1.000	1.000
EC	0.7 C	1.000	0.939	0.983	1.000	0.984	1.000	0.990	0.966	1.000	1.000
EC	1.4 C	0.001	0.001	0.927	0.000	0.433	0.001	0.000	0.000	0.003	0.129

Legend: EO – eyes open; EC – Eyes closed.

with the scenery oscillating at frequencies of 0.2 Hz and 0.5 Hz were compared with the measurements taken with the "closed" scenery oscillating at a frequency of 1.4 Hz. Tables 7–9 contain the calculated CVs obtained for individual measurements. It is also worth noting that the values of these coefficients obtained for changes in the trend are smaller in the case of the number of changes in the trend than in the case of PSD, especially in the case of high frequencies.

Table 6. *p* values ($\alpha = 0.05$) for the comparison of PSD results between studies in real and virtual environments

Research	Time interval [s]	0.05–0.1	0.1–0.2	0.2–0.3	0.3–0.4	0.4–0.5	0.5–0.6	0.6–0.7	0.7–0.8	0.8–0.9	0.9–1.0
EO	EC	0.005	0.018	0.001	0.011	0.018	0.010	0.754	0.998	0.483	0.011
EO	0.2 O	0.056	0.280	0.006	0.010	0.027	0.217	0.208	0.106	0.064	0.003
EO	0.5 O	0.003	0.004	0.001	0.000	0.000	0.002	0.021	0.386	0.061	0.000
EO	0.7 O	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.101	0.026
EO	1.4 O	0.000	0.000	0.001	0.000	0.052	0.109	0.648	0.994	0.885	0.300
EO	0.2 C	0.000	0.000	0.000	0.000	0.000	0.000	0.027	0.017	0.002	0.005
EO	0.5 C	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000
EO	0.7 C	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.045	0.079
EO	1.4 C	0.000	0.000	0.000	0.000	0.000	0.000	0.015	0.008	0.006	0.292
EC	0.2 O	1.000	1.000	0.996	0.967	0.989	1.000	0.913	0.363	0.826	0.861
EC	0.5 O	0.928	0.868	0.915	0.286	0.288	0.809	0.419	0.770	0.817	0.008
EC	0.7 O	0.121	0.033	0.058	0.017	0.007	0.064	0.000	0.000	0.796	0.948
EC	1.4 O	0.387	0.170	0.784	0.000	0.974	0.998	0.995	1.000	1.000	1.000
EC	0.2 C	0.850	0.242	0.449	0.298	0.144	0.751	0.577	0.116	0.320	0.983
EC	0.5 C	0.244	0.003	0.005	0.003	0.006	0.071	0.000	0.002	0.024	0.000
EC	0.7 C	0.007	0.003	0.001	0.011	0.006	0.009	0.000	0.000	0.483	0.961
EC	1.4 C	0.000	0.000	0.001	0.000	0.003	0.014	0.182	0.032	0.161	0.999

Table 7. Coefficient of variation in individual measurements in the real environment

CV \ Time interval [s]	0.05–0.1	0.1–0.2	0.2–0.3	0.3–0.4	0.4–0.5	0.5–0.6	0.6–0.7	0.7–0.8	0.8–0.9	0.9–1.0
eyes open										
PSD	30.1	52.6	68.4	58.9	67.1	62.5	88.8	51.7	73.6	92.1
TCI	27.8	22.7	17.6	17.6	22.7	21.4	25.0	50.0	100.0	–
eyes closed										
PSD	41.9	55.9	65.0	64.3	62.2	75.2	90.9	125.1	86.3	100.5
TCI	25.0	26.3	18.8	19.4	16.7	21.4	25.0	50.0	50.0	

Table 8. Coefficient of variation in individual measurements in the virtual environment in "open" scenery

CV \ Time interval [s]	0.05–0.1	0.1–0.2	0.2–0.3	0.3–0.4	0.4–0.5	0.5–0.6	0.6–0.7	0.7–0.8	0.8–0.9	0.9–1.0
open scenery 0.2 Hz										
PSD	24.3	48.8	48.8	44.8	64.8	51.2	88.7	105.0	89.2	72.2
TCI	28.6	25.8	25.0	11.8	22.7	28.3	46.4	50.0	100.0	
open scenery 0.5 Hz										
PSD	33.3	49.0	65.6	120.7	108.5	82.9	112.3	99.0	108.0	85.1
TCI	30.8	24.2	22.4	25.0	18.5	30.4	30.0	31.3	50.0	100.0
open scenery 0.7 Hz										
PSD	73.9	51.6	110.7	99.7	87.4	59.8	137.9	89.7	57.7	47.4
TCI	50.0	42.3	16.1	23.5	13.6	18.8	16.7	100.0	50.0	
open scenery 1.4 Hz										
PSD	20.7	64.5	62.0	62.2	60.4	41.5	66.6	71.7	94.7	94.1
TCI	38.9	72.9	22.6	18.4	24.1	54.2	75.0			

Table 9. Coefficient of variation in individual measurements in the virtual environment in "closed" scenery

CV \ Time interval [s]	0.05–0.1	0.1–0.2	0.2–0.3	0.3–0.4	0.4–0.5	0.5–0.6	0.6–0.7	0.7–0.8	0.8–0.9	0.9–1.0
closed scenery 0.2 Hz										
PSD	43.6	116.7	111.8	124.7	63.9	104.7	89.3	101.7	54.7	73.1
TCI	28.6	15.8	21.1	18.4	18.2	21.4	25.0	25.0	50.0	
closed scenery 0.5 Hz										
PSD	62.5	66.8	93.4	123.8	89.5	75.5	150.6	70.9	106.9	76.5
TCI	40.0	16.7	18.4	20.6	13.6	33.3	37.5	50.0	100.0	
closed scenery 0.7 Hz										
PSD	233.2	430.4	202.8	337.4	202.0	167.6	143.2	112.7	81.1	52.0
TCI	25.0	23.5	28.9	11.5	23.8	16.1	37.5	25.0	75.0	
closed scenery 1.4 Hz										
PSD	96.2	124.9	121.7	213.7	343.8	142.4	193.2	214.3	242.7	397.9
TCI	18.8	41.7	42.3	25.0	16.7	33.3				

4. Discussion

The analysis of successive positions of the COP in terms of the path length it traveled during the measurement, the average speed with which it was moving, the area in which it was located or the range of displacements along a specific direction are the simplest and most common methods of assessing the ability to maintain balance. In a person with balance dysfunctionalities these values increase, which, in turn, allows for a preliminary diagnosis of problems with balance or the assessment of progress in rehabilitation. However, these approaches do not always work well. Use of an unstable surface or conditions causing sensory conflict cause all these values to increase in healthy people. In such cases, analysis in the frequency domain becomes helpful, as it enables the decomposition of the COP movement signal into cyclic components and thus allows for the determination of the frequen-

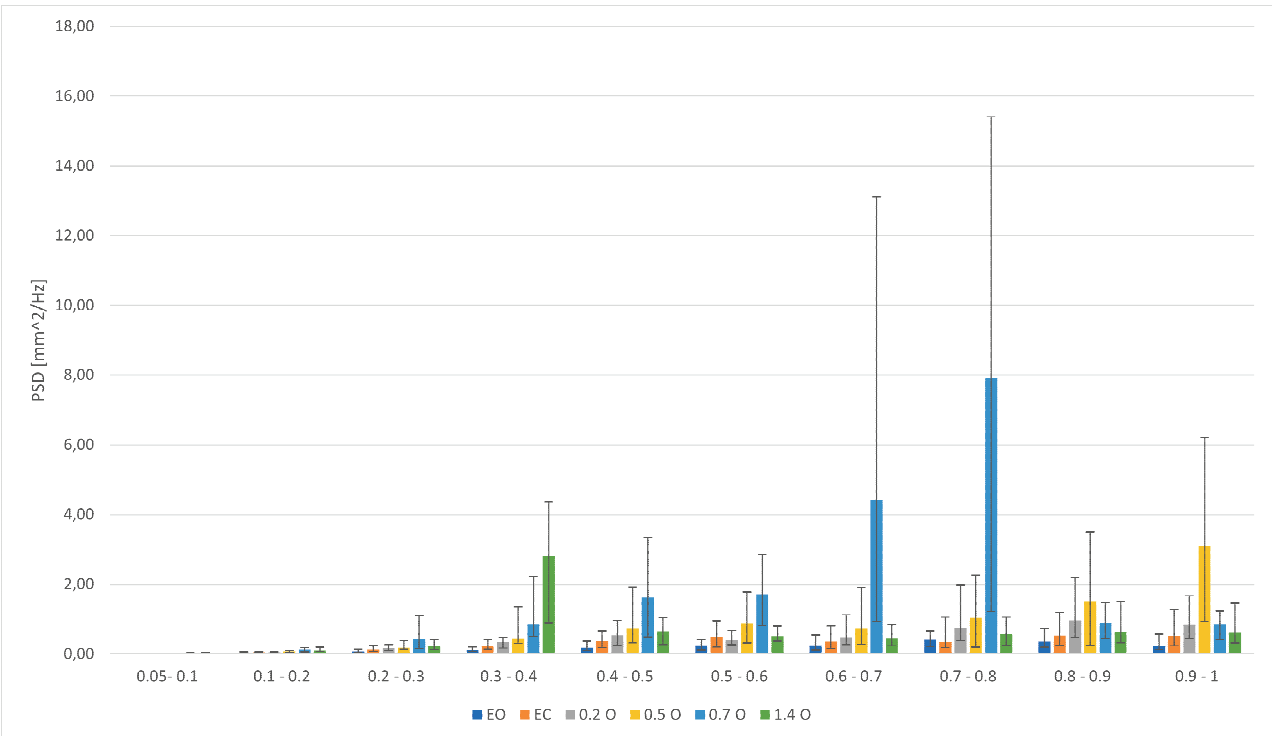


Fig. 7. PSD values obtained for individual time intervals for measurements in the real environment and in the open scenery

cies of the main movement components. In this way, components of movement such as the body following a cyclically moving environment become visible, which, in turn, increases the possibilities for interpreting the received signal [12], [21]. However, the observation of successive positions of the COP indicates that the continuous correction of the position, velocity and acceleration of the COM, and the resulting changes in the position of the COP may not only be cyclical but also random. Both traditional time domain methods and frequency domain analysis cannot detect such non-cyclic corrections. The technical analyzes proposed in this paper for determining the trend changes in currency exchange rates and stock market shares made it possible to detect the trend changes occurring in the COP movement (i.e., significant changes in the direction of the movement of this point), the nature of which was both cyclical and non-cyclical. As part of

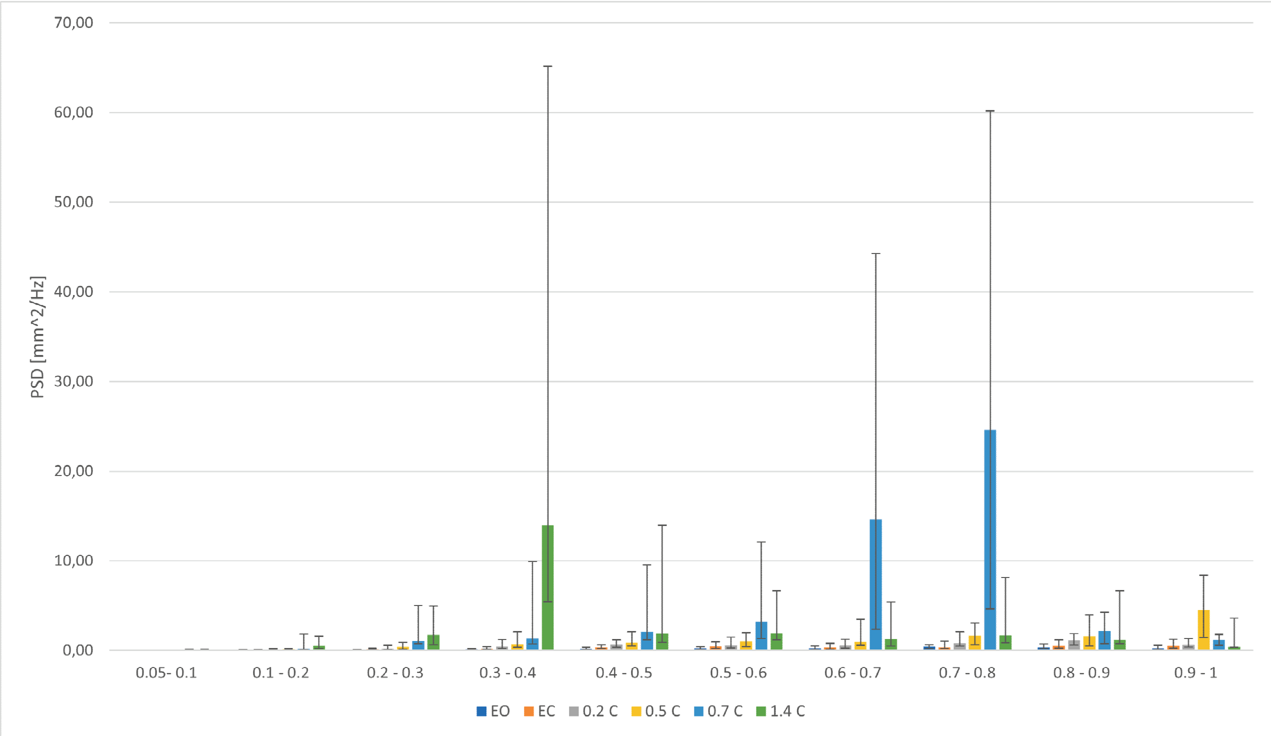


Fig. 8. PSD values obtained for individual time intervals for measurements in the real environment and in a closed scenery

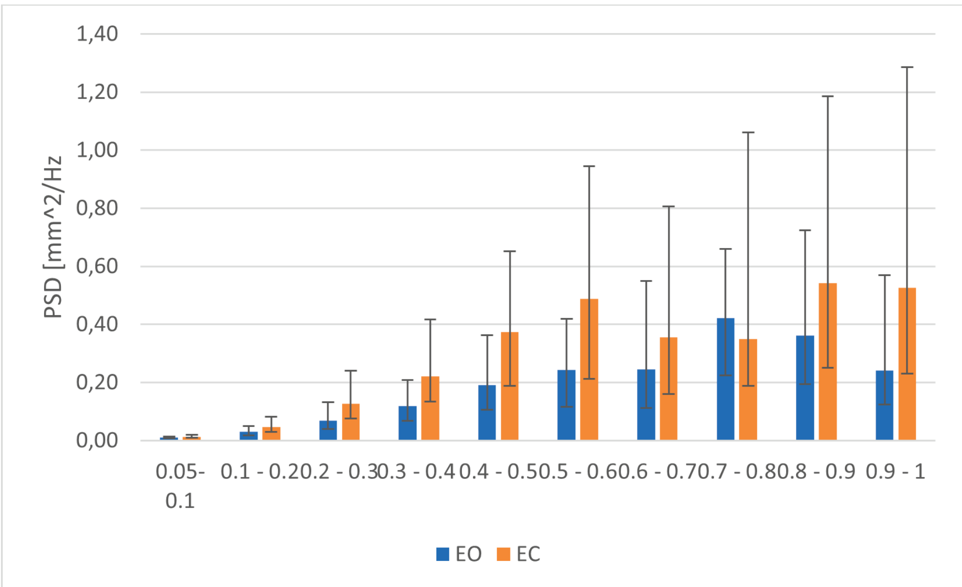


Fig. 9. PSD values obtained for individual time intervals for measurements in real environment

the work, comparative analyzes were carried out in the field of time and frequency using calculations used for stock market analyzes in order to present the interpretative possibilities of these approaches in the assessment of the ability to maintain balance. Calculations for a new coefficient that defines the number of trend changes in the COP displacement signal were proposed.

Analysis of individual measurements

The analysis of the results obtained in the time domain shows that the introduction of oscillating scenery increases the values of the analyzed quantities in the real environment. Both the speed and the range of motion increased, however, these values do not make it possible to indicate the mechanism of these changes (Table 2).

PSD is used to indicate the dominant frequencies in the analyzed signal, especially in the cases where a large amount of noise is present. The PSD presented in Fig. 4a shows values below 0.5 Hz and above 0.5 Hz and indicates that the majority of cyclic components appearing in the analyzed signal have a frequency lower than 0.5 Hz. The analysis of *p* coefficients (Table 3) showed no statistically significant differences with the scenery oscillating at a frequency of 0.7 Hz for both types of scenery and for "closed" scenery with an oscillation frequency of 1.4 Hz. This indicates that, for these cases, the values of the COP motion amplitude resulting from the body following the moving scenery were so large that they were visible in the frequency range above 0.5 Hz. This result seems to be consistent with the conclusions of other studies [12], [21]. Such a distribution of cyclic components may indicate the dominance of visuo-vestibular control [2], [15], when the body performs mainly slow, cyclic movements while standing. The increases in the PSD value in the case of research in the virtual environment may indicate the appearance of additional cyclical movements related to the oscillation of the scenery. These changes include both increases at frequencies lower than 0.5 Hz and higher than 0.5 Hz – the latter case is particularly visible in measurements conducted in "closed" scenery with the oscillation frequencies of this scenery equal to 0.7 Hz and 1.4 Hz. Analysis of the data presented in Figure 8 shows that this increase is mainly caused by increases in the PSD value in the time interval of 0.3–0.4 s for the test with a frequency of 1.4 Hz and the time interval of 0.6–0.8 s for the test with a frequency of 0.7 Hz. Such PSD may show that the "closed" scenery together with the higher oscillation frequency had a greater influence on the behavior of the participants, causing them to have significant cyclic movements synchronized with the oscillation of the scenery.

The analysis of the sum of individual trend changes for individual measurements (Fig. 4b) shows that the number of these changes does not depend on the conditions. This may mean that balance control requires a certain number of rapid changes in the trend of the COP movement (for the time interval of 0.05–1.0 s, corresponding to frequencies of 10 Hz to 0.5 Hz). Making the conditions less favorable, which can include the introduction of a disturbance in the form of oscillating scenery, does not significantly change this number. On this basis, it can be concluded that either the balance system does not need more of this type of movement or the locomotor system is not able to generate more of them.

One more conclusion can be drawn from the PSD analysis and the sum of trend changes. The values of the CVs (Table 4) obtained for PSD only in the case of measurements with EC and for measurements at a frequency of 0.7 Hz (in both cases for a result below 0.5 Hz) take a value lower than 40%. In the rest of cases, they are close to or much greater than 50%. This demonstrates a large dispersion of the obtained measurement data. In the case of the number of changes in the trend, these values exceed 10% only in one case. This may indicate that in this type of analysis, when it is important to study insignificant but rapid movements of the COP, the use of analyzes based on stock exchange ratios may bring much better results.

Analysis with division into time intervals

Observed signal power in the range above 0.5 Hz obtained in tests performed in the real environment (Fig. 9), as well as in tests in the virtual environment with the scenery oscillating at a frequency of 0.2 Hz and 0.5 Hz (Figs. 7, 8), is negligibly small and does not exceed most of the analyzed unity intervals. The exception is the range of 0.8–1.0 s, where slightly higher PSD values obtained during tests at a frequency of 0.5 Hz were observed. This can be explained by the fact that the time interval of 0.8–1.0 s corresponds approximately to this frequency (i.e., the movements of the body following the scenery were observed here).

In the case of frequencies of 0.7 Hz and 1.4 Hz, a significant increase in the PSD value was observed in the time intervals corresponding to these frequencies (0.3–0.4 s for the frequency of 1.4 Hz and 0.6–0.8 s for the frequency of 0.7 Hz). These increases indicate that the COP, and thus the entire body, followed the cyclically moving scenery. The differences in the PSD values visible between the "open" and "closed" scenery result from differences in the amplitude values of the recorded movements following the scenery.

The analysis of the number of trend changes obtained during the tests in the real environment (Figs. 5, 6) shows that in the intervals from 0.05 s to 0.5 s (which corresponded to the frequency range of 0.5–10 Hz movements) there was a significant number of changes in the trend indicating changes in the direction of the COP movement; this was despite very low PSD values. According to the information provided by Bizid et al. [2] and Micarelli et al. [15], this may indicate an important role of proprioceptive control, despite the PSD analysis indicating a lack of cyclic movements in this frequency range and suggesting that there is a minor role for this type of control.

Interesting conclusions are provided by analysis of the number of trend changes in subsequent time intervals obtained in the virtual environment in relation to results obtained in the real environment (Figs. 5, 6). In the case of scenery oscillation frequencies of 0.7 Hz and 1.4 Hz for the time interval of 0.05–0.2 s, most cases saw a statistically significant decrease in the number of trend changes observed compared to the measurements with EO and EC, with a simultaneous increase of this number in the time intervals corresponding to the scenery oscillation frequencies (0.3–0.5 s). For the frequency of 1.4 Hz, a drop in the number of trend changes was also recorded for the time interval of 0.5–0.9 s. This means that the increased number of trend changes appearing at the frequencies of 0.7 Hz and 1.4 Hz reduced the number of trend changes at other frequencies. When analyzing the possible causes of these changes, it is necessary to return to the previous point that stated the total number of changes in the trend for the 0.05–1.0 s range undergoes only very slight changes, regardless the measurement carried out (Fig. 4b). This may mean that the introduction of disturbances with a frequency in this range (e.g., 1.4 Hz) resulted in an increase in the number of body movements performed with this frequency (in an attempt to follow the scenery to prevent a potential fall) and forced a reduction in the number of trend changes in other time periods, whilst the total the number of all trend changes remained approximately constant. It can be assumed that this invariability in the maximum number of trend changes may be related to the fact that the locomotor system is not able to increase the total number of trend changes above a certain value. In such case, introducing environmental oscillations with a specific frequency may reduce the number of trend changes in the range of 0.05–0.2 s, where these fast movements of COP are treated as related to proprioception. This reduction in the number of changes may not allow the proprioceptive system to obtain sufficient information and thus the individual may not be able to maintain their balance. This conclusion shows that cyclical oscillations of the environment not only affects the information collected by the visuo-vestibular system, but may also distort information from the proprioceptive system.

Additionally, it should be noted that the coefficients of variation obtained for the trend changes are much smaller than those obtained for the PSD (Tables 4–9), which may indicate that the analysis of these values provides much more reliable results. Based on the above considerations, it can be assumed that:

- reducing the number of trend changes in the frequency band assumed to be used in proprioceptive control may have an impact on the ability to maintain balance; this information and the practical use of stock exchange ratios can be helpful in the diagnosis of people with balance dysfunctionalities;
- by analyzing changes in the number of changes in the trend in individual time intervals, it is possible to selectively assess the impact of proprioceptive control on the overall ability to maintain balance.

Limitations

The presented method for assessing the ability to keep balance is an extension of the methods used so far, however, it also has some limitations and elements that require further research, including:

- stating the extent to which changes in the parameters used for the calculations of TCI will affect the final results;
- determining whether the application of this method will give comparable or more reliable results in the case of using parameters that enable the analysis of slow changes in the trend (below 0.5 Hz) compared to the analyzes carried out in the frequency domain (i.e., Fourier transform, wavelet analysis);
- checking whether a decrease in the number of changes in the trend in the range of 0.05–0.2 s (above 0.5 Hz) has a negative impact on the ability to maintain balance;
- investigating whether smaller fluctuations in the number of changes in the trend mean less impact of the oscillating environment on the individual.

5. Conclusions

Summing up, it can be stated that the application of approaches for determining stock exchange indices in the analysis of subsequent COP positions provides new information that can be used in the assessment of the ability to maintain balance. This knowledge can be used both in clinical studies on balance and in the development of new methods using this type of measurement. The values of the proposed index may be used to determine the excitation of systems responsible for maintaining balance, as they constitute an objective tool for determining the number of COP displacement corrections per unit of time. These types of indicators may be appropriate for clinical applications.

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applied sciences

Article

Impact of Visual Disturbances on the Trend Changes of COP Displacement Courses Using Stock Exchange Indices

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Abstract: This work aims to define a strategy for maintaining a vertical posture of the human body under conditions of conflicting sensory stimuli using a method of trend change analysis. The investigations involved 28 healthy individuals (13 females, 15 males, average age = 21, SD = 1.3 years). Measurements were conducted with eyes opened and closed and in the virtual environment with two sceneries oscillating at two frequencies. Values in the time domain were calculated—the mean center of pressure (COP) velocity and movement range in the AP direction—as well as values based on the moving average convergence divergence (MACD) computational algorithm—the trend change index (TCI), MACD_dT, MACD_dS, and MACD_dV. After dividing the analysis into distinct time periods, an increase in TCI values was identified in the oscillating scenery at 0.7 and 1.4 Hz during the 0.5–1 and 0.2–0.5 s time periods, respectively. Statistically significant differences were observed between measurements with an oscillation frequency of 0.7 Hz and those with an oscillation frequency of 1.4 Hz during the 0.2–0.5 s and 0.5–1 s periods. The use of stock exchange indices in the assessment of the ability to keep a stable body posture supplements and extends standard analyses in the time and frequency domains.

Keywords: virtual reality; center of pressure; postural stability; stock market indicators

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1. Introduction

Maintaining a stable body posture results from the constant control and corrections performed by many cooperating systems, including the vestibular system and deep sensibility [1]. The nervous system constantly stimulates muscles that compensate for momentary disturbances and maintains the center of mass within the boundaries of stability [2,3]. Posture stability tests are devised to determine values based on the coordinates of the subsequent center of foot pressure (COP) locations. Commonly analyzed values that describe the ability to maintain balance include COP velocity, COP ellipse surface area, and COP movement ranges in the antero-posterior (AP) and medio-lateral (ML) directions. Analyzing the above-mentioned values, it is assumed that the higher these values, the less likely that one can maintain a stable body posture [4]. The above-mentioned analyses in the time domain are supplemented with analyses in the frequency domain. Frequency analyses based on calculations of the fast Fourier transform (FFT) make it possible to determine the cyclic components of the analyzed signal and ascertain whether a cyclic stimulus upsetting balance translates into the cyclicity of COP or center of mass (COM) movements [5,6]. Such analysis also finds application in the case of tests performed under conditions of conflicting sensory stimuli. In such situations, individual senses receive contradictory information and a given person (test participant) is affected by stimuli—both in the real-life environment and virtual reality—which, directly by physical push or indirectly by visual stimuli, unbalance the person [1,2,7,8]. Research in the virtual environment can be mentioned as an example of measurements under sensory conflict conditions where oscillating virtual scenery is used with the still, real floor [1]. Relevant literature also describes other methods to determine the ability of the human body to maintain balance. Popular methods include probabilistic analyses based on entropy [9] and frequency analyses using the wavelet distribution [10,11]. The latter is specifically applied to the analyses of low frequencies of COP movements.

The choice of analysis tools depends on the conditions of the conducted tests and the information researchers require from the conducted tests. Analyses in the time and frequency domains enable a complex and full evaluation of the ability to keep balance [12,13]. Frequency domain analyses are a natural extension of time domain analyses. The frequency analyses described in scientific publications usually determine the impact of cyclic external disturbances on the ability to maintain balance by checking if cyclic components are part of the movement and whether the following scenery is a dominant one [1,5]. To deepen the analyses related to the determination of the cyclicity of COP movements, it is necessary to use a method that would detect momentary, non-cyclic changes occurring during the whole analysis [14,15]. Such a possibility is provided by trend change analysis making use of stock exchange indices [16,17]. The method of stock exchange indices may be used both under static and dynamic conditions and, as in the analysis of COP movements, during which the patient wears a virtual reality (VR) projection headset. The proposed methodology for the analysis of trend changes based on determining the changing points of the COP displacement signal trend allows us to supplement traditional analysis methods with additional information related to determining the strategy of maintaining balance by a human. The algorithm for determining trend change points is based on time domain analysis, so the results are not as noisy as in the case of FFT analyses and spectral leakage. This procedure enables us, for a certain frequency range, to detect both cyclical components and non-cyclical changes [16].

The objective of this work is to develop a method of analysis for maintaining the vertical position of the human body under the conditions of conflicting sensory stimuli using an innovative method of trend change analysis. This work is a continuation of previous research [16] regarding the assessment of human balance under conditions of conflicting sensory stimuli—oscillating virtual scenery and stable floor.

This paper investigates the application of stock exchange indices connected with the evaluation of stock rate courses to study the ability of the human body to maintain balance in an oscillating virtual reality. The performed analyses may confirm and extend the usability of stock exchange indices as a tool to supplement standard analyses for balance assessments [18].

To maintain a stable body posture, it is necessary to constantly correct COP and center of mass (COM) positions. Analysis using stock exchange indices will allow us to determine the number of posture corrections, the time and distance between these corrections, and the speed of movement of the COP between these corrections [19]. We hypothesize that a decrease in the frequency of posture corrections, coupled with an increase in the distance between subsequent points of the trend change, may indicate an increased risk of falling. The proposed method, based on the analysis of posture corrections, will allow for the detection of the impact of changes in the frequency of visual disturbances or different sceneries on postural changes. If validated, this method could effectively assess temporary postural changes. We also assume that changes in the total number of trend changes and in particular time periods may indicate dysfunctions that have an impact on the balance-keeping ability.

2. Materials and Methods

2.1. Measurement Stand

The measurement stand consisted of a stabilographic platform (WinFDM-S, Zebris, Isny im Allgau, Germany, 100 Hz sample frequency, 2560 tensiometer sensors, 34 cm × 54 cm sensing area) and a head-mounted display (HMD) Oculus Rift set used for the projection of three-dimensional images. The three-dimensional sceneries were developed in the Unity 3D environment (Figure 1A,B).

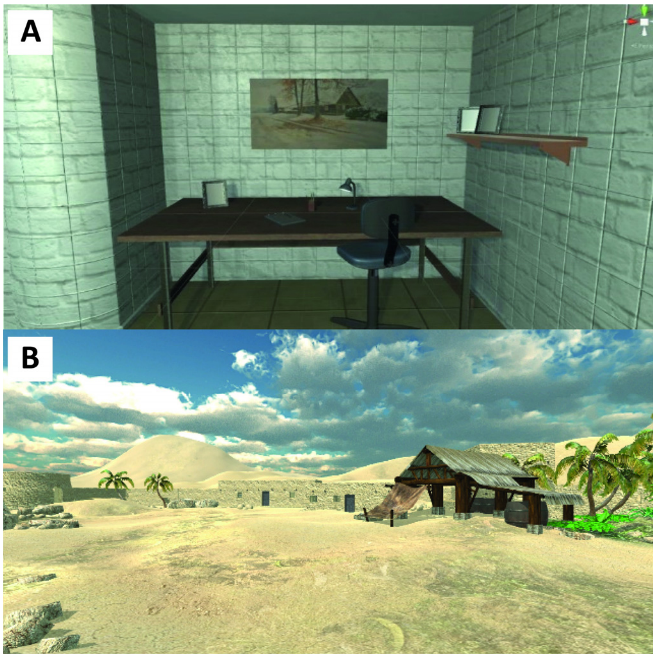

Figure 1. (**A**,**B**) Scenery used in the tests. (**A**) closed space, (**B**) opened space.

The 'closed space' scenery (Figure 1A) consisted of a furnished room, in which objects were seen by the tested individual at ~3 m, whereas the 'open space' scenery (Figure 1B) showed a desert, where objects were located at ~100 m. During the tests, the scenery oscillated in the AP direction at a constant frequency. Oscillations were presented by moving the scenery in the AP direction, taking into consideration a slight rotation of the scene by 0.5 degrees [1]. The scenery oscillations were perceived by the subject as the movement of the entire environment. The oscillation amplitude was set at 30 cm, whereas the oscillation frequency was set following the procedure described below.

2.2. Study Group

The investigations involved 28 individuals (13 females and 15 males) with an average age of 22 years (1.3 standard deviation (SD)), mean body mass of 67.5 kg (12 SD), and an average height of 173.6 cm (8.8 SD). All participants took part in the measurements conducted in a real-life environment. A group of 12 individuals (7 females and 5 males, average age 22.5 years (0.5 SD), mean body mass 65.1 kg (13.1 SD), average height 171.9 cm (8.8 SD)) was subjected to measurements in the VR 'closed space' scenery, while 16 participants (6 females and 10 males, average age 21.7 years (1.6 SD), mean body mass 69.3 kg (10.8 SD), average height 174.9 cm (7.3 SD)) were tested in the VR 'open space' scenery. Health problems related to maintaining balance, for example, neurodegenerative diseases or labyrinth problems, and obesity (body mass index; BMI > 30) were considered as exclusion criteria. This study was first approved by the Ethics in Research Committee of the Academy of Physical Education in Katowice (protocol number 5/2020). Each participant gave verbal consent to the measurements.

2.3. Experimental Procedure

The experimental procedure encompassed tests in the real-life environment, involving standing with opened eyes (OE) and standing with closed eyes (CE), as well as tests in the virtual environment (Figure 2).

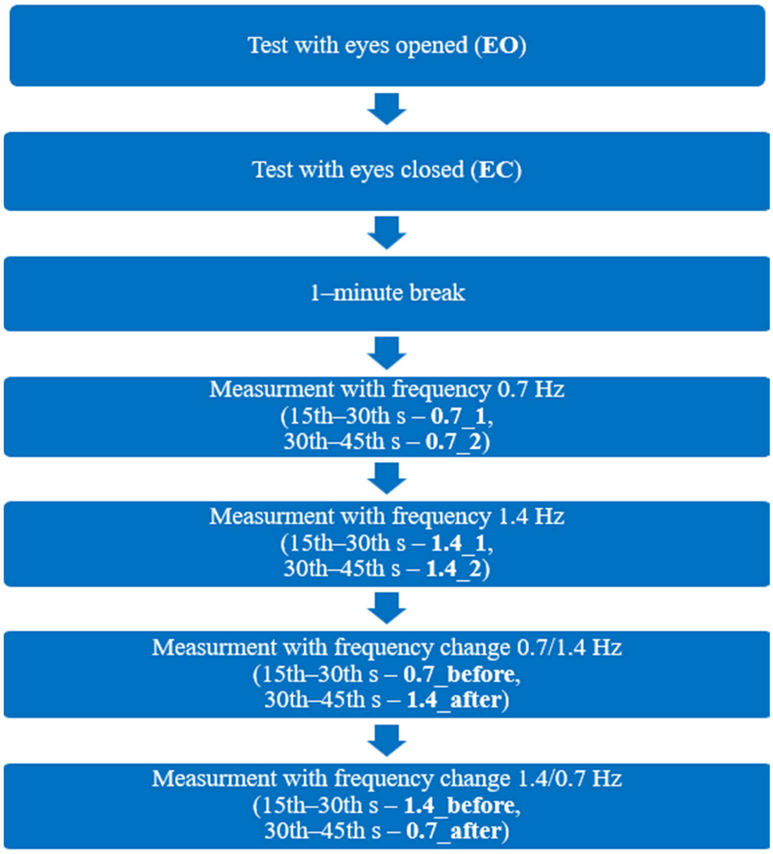

Figure 2. Experimental procedure.

The measurements in VR were conducted by presenting the open space scenery and closed space scenery oscillating at frequencies of 0.7 Hz and 1.4 Hz. The selection of frequency values was based on previously conducted research, which proved that during the application of such values and the HMD headset, the oscillating scenery had a strong impact on the tested individuals, such as the movement of following the scenery and increasing the values of basic stabilographic quantities [1,16]. The subjects were asked to take off their shoes and stand on the measurement platform in an upright position with their arms crossed on their chest and their vision focused on a point in the form of a dot on the wall located in front of them. The subjects' task was to simply stand still during all measurements. First, measurements in the real-life environment were performed. Then, measurements were taken in the virtual environment while wearing 3D goggles. Measurements in the virtual environment took place directly one after another, without breaks. A measurement in the oscillating scenery at a frequency of 0.7 Hz (0.7), a measurement in the oscillating scenery at a frequency of 1.4 Hz (1.4), a measurement in the oscillating scenery with the change in oscillation frequency from 0.7 Hz to 1.4 Hz (0.7/1.4), and a measurement in the oscillating scenery with the change in oscillation frequency from 1.4 Hz to 0.7 Hz (1.4/0.7) were all taken (Figure 2). The measurement series was performed once for each participant. The change in the oscillation frequency occurred at the half-time of the measurement. Each measurement lasted 60 s. In the measurements with the oscillating scenery, the movement of the scenery began in the 15th s of the measurement and lasted 30 s. All analyses took into consideration a period from the 15th to the 45th s, in compliance with the description contained in the chapter 'Analyzed Quantities'.

2.4. Analyzed Quantities

The analysis involved displacements of the COP in the AP direction at the time of the middle 30 s of the test with the frequency change (15 s before and 15 s after the change), obtaining the courses of 0.7_before and 1.4_after from the 0.7/1.4 measurement as well as courses 1.4_before and 0.7_after from the 1.4/0.7 measurement. In the case of the OE and CE measurements, the time of the middle 15 s was analyzed. Conversely, the 0.7 and 1.4 measurements used the first 15 s (0.7_1, 1.4_1) and the second 15 s (0.7_2, 1.4_2) of the test for analysis. The analyses were performed using the MATLAB R2021b software program. The signal from the platform was filtered with a Butterworth low-pass filter. The pass frequency of the filter was set to 7 Hz. Basic stabilographic values were determined, namely the COP velocity and the COP movement range in the AP direction. For each course, the trend change index (TCI) was calculated [18,20,21]. The TCI defines the number of trend changes, which is the number of intersections of the signal resulting from the computations of the algorithm: moving average convergence divergence (MACD) (Equations (1)–(3)).

To calculate TCI values, the MACD line for the COP signal was determined by computing the difference between two exponential moving averages (EMAs) with lengths of 12 and 26 samples, as per Equations (1) and (2):

$$\mathrm{MACD} = \mathrm{EMA}_{\mathrm{COP},12} - \mathrm{EMA}_{\mathrm{COP},26} \tag{1}$$

where $\mathrm{EMA}_{\mathrm{COP},12}$—faster exponential moving average for COP signal, and $\mathrm{EMA}_{\mathrm{COP},26}$—slower exponential moving average for COP signal.

$$\mathrm{EMA} = (p_0 + (1-\alpha)p_1 + (1-\alpha)^2 p_2 + \cdots + (1-\alpha)^N p_N)/(1 + (1-\alpha) + (1-\alpha)^2 + \cdots + (1-\alpha)^N) \tag{2}$$

where p_0—ultimate value, p_1—penultimate value, p_N—value preceding N periods, N = number of periods, and α = a smoothing coefficient equal to 2/(N + 1).

Moving on to the subsequent phase, the signal line was calculated as an EMA with a length of nine samples from the MACD line signal, in accordance with Equation (3):

$$\text{Signal line} = \mathrm{EMA}_{\mathrm{MACD\ line},9} \tag{3}$$

The TCI was presented as the total number of all trend changes for the analyzed measurement and the total number of trend changes segmented into the following time periods:

- 0–0.2 s,
- 0.2–0.5 s,
- and 0.5–1 s.

These time periods denote the times after which the subsequent trend change occurred after the previous one. Based on the MACD algorithm, the following values were also computed: the mean distance between subsequent points of the trend change (MACD_dS), the mean time between subsequent points of the trend change (MACD_dT) (Figure 3), and the mean velocity of changes of displacements between subsequent points signifying trend changes (MACD_dV).

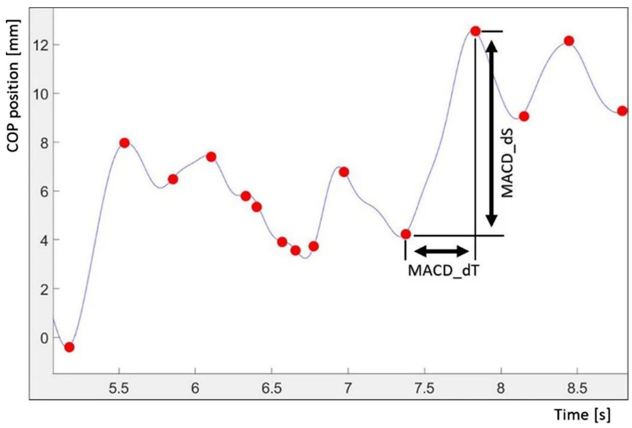

Figure 3. COP course with trend change moments detected by the algorithm (marked as red dots in the diagram) and graphic interpretation of MACD_dT and MACD_dS for two random trend change moments [21].

2.5. Statistical Analysis

Statistical analysis was performed using the MATLAB R2021b software program. The compared values were evaluated using the Shapiro–Wilk test to check for the occurrence of normality in the distribution. The results did not confirm the occurrence of normal distributions in all sub-groups. Therefore, medians are presented in the diagrams and non-parametric tests were used.

In the first stage of comparative analysis, comparisons were made between the results calculated for the group of people assessed in the closed scenery and the group of people tested in the open scenery. The Mann–Whitney U test was used for the analysis, considering different group sizes. Comparisons were made for all calculated values separately and no statistically significant differences were found ($p < 0.05$, effect size d > 0.7).

Due to the lack of significant differences, it was decided to combine the groups into one. For all persons, the following values were determined: COP velocity, COP movement, range of COP motion in the AP direction, TCI, MACD_dT, MACD_dS, and MACD_dV, and then for all parameters, the differences between subsequent tests were examined (OE, CE, 0.7_1, 0.7_2, 1.4_1, 1.4_2, 0.7_before, 0.7_after, 1.4_before, and 1.4_after). To verify whether there were statistically significant differences in the case of the analyzed groups, the ANOVA Friedman test was conducted followed by the pairwise Wilcoxon post hoc test with Holm correction. All performed tests obtained high test power.

3. Results

The values were obtained based on the algorithm calculating the MACD index. No statistically significant differences were found in any groups for all of the analyzed values between the measurements of open and closed space sceneries. Therefore, the obtained values were merged into one group. The results were divided into three groups: standard values in the time domain and values computed based on the MACD algorithm, including MACD_dT, MACD_dS, MACD_dV, and TCI, with division into time periods.

3.1. Data Analysis in Time Domain

The diagrams below (Figure 4A,B) present the medians of the stabilographic values in the time domain—the mean COP velocity in the AP direction and the range of COP movement in the AP direction.

The values of the COP velocity in the AP direction (Figure 4A) increased statistically significantly after a participant closed their eyes and after the introduction of disturbances in VR. Statistically significant differences were present while comparing the OE measurements with measurements using VR and while comparing the CE measurements with tests in the virtual environment, except for 1.4_2, 0.7_before, and 1.4_before. No statistically significant differences occurred in the comparison of values obtained during the measurements conducted in VR. The diagram (Figure 4B) shows an increase in the values of the COP movement range in the AP direction with the application of disturbances in VR in comparison with the values obtained in the real-life environment. An increase occurred in the case of the median values, interquartile distribution, and data distribution. In the case of the median value, statistically significant differences were observed between the OE measurements and the measurements performed in VR, except for 1.4_2, 0.7_before, and 1.4_before. An analogous situation occurred between the CE measurements and measurements with the oscillating scenery 0.7_1.4 and 1.4_1.

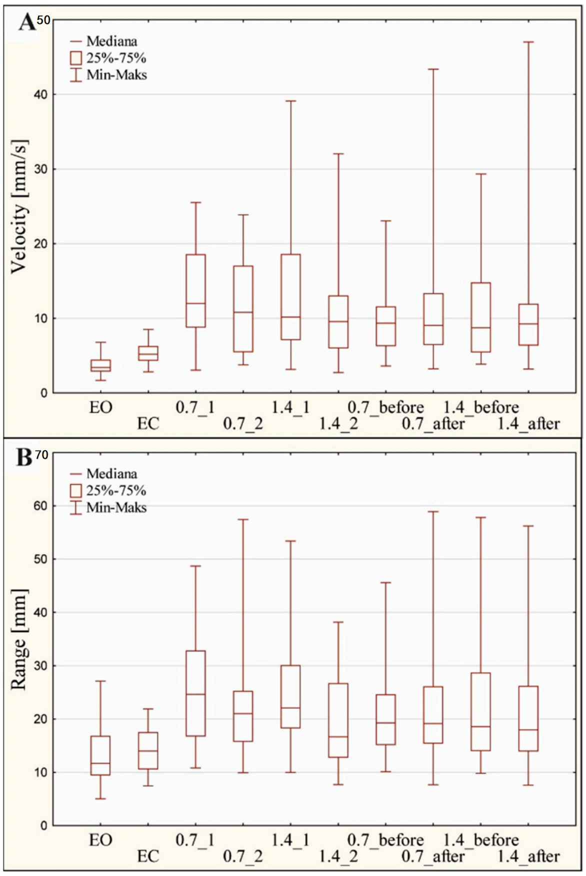

Figure 4. (**A**,**B**). Stabilographic values in the time domain (EO—eyes opened, EC—eyes closed, 0.7_1—first 15 s of measurement in the oscillating scenery at a frequency of 0.7 Hz, 0.7_2—second 15 s of measurement in the oscillating scenery at a frequency of 0.7 Hz, 1.4_1 first 15 s of measurement in the oscillating scenery at a frequency of 1.4 Hz, 1.4_2 second 15 s of measurement in the oscillating scenery at a frequency of 1.4 Hz, 0.7_before—first 15 s of measurement in the oscillating scenery with the change in oscillation frequency from 0.7 Hz to 1.4 H, 1,4_after—second 15 s of measurement in the oscillating scenery with the change in oscillation frequency from 0.7 Hz to 1.4 Hz, 1.4_before—first 15 s of measurement in the oscillating scenery with the change in oscillation frequency from 1.4 Hz to 0.7 Hz, and 0.7_after—second 15 s of measurement in the oscillating scenery with the change in oscillation frequency from 1.4 Hz to 0.7 Hz). (**A**) Median of mean COP velocity in the AP direction and (**B**) Range of COP movement in the AP direction.

3.2. Trend Change Index

The diagrams below (Figure 5A–D) present the medians of the TCI value. Table 1 presents the *p*-values obtained for the comparison of the TCI values.

In the diagram presenting the TCI values for individual measurements (Figure 5A), one can observe that the TCI value obtained for the measurements using the oscillating scenery at a frequency of 1.4 Hz was at a similar level to the OE and CE measurements. The TCI values in the case of disturbances at a frequency of 0.7 Hz were lower than the values obtained in other measurements. The data distribution was also noticeably higher. The comparison of the OE and CE measurements did not reveal any statistically significant differences. Statistically significant differences were observed when comparing the OE test in a real-life environment to measurements in the virtual environment for the case of the oscillating scenery at a frequency of 0.7 Hz—both before (0.7_before) and after (0.7_after) the change in frequency. However, in the case of the CE test, statistically significant differences did not occur except for the oscillating scenery experiment at a frequency of 0.7 Hz after the change in frequency (0.7_after). When comparing the tests conducted in VR, statistically significant differences were found only in experiments where the frequency of the oscillating scenery changed from 1.4 Hz to 0.7 Hz.

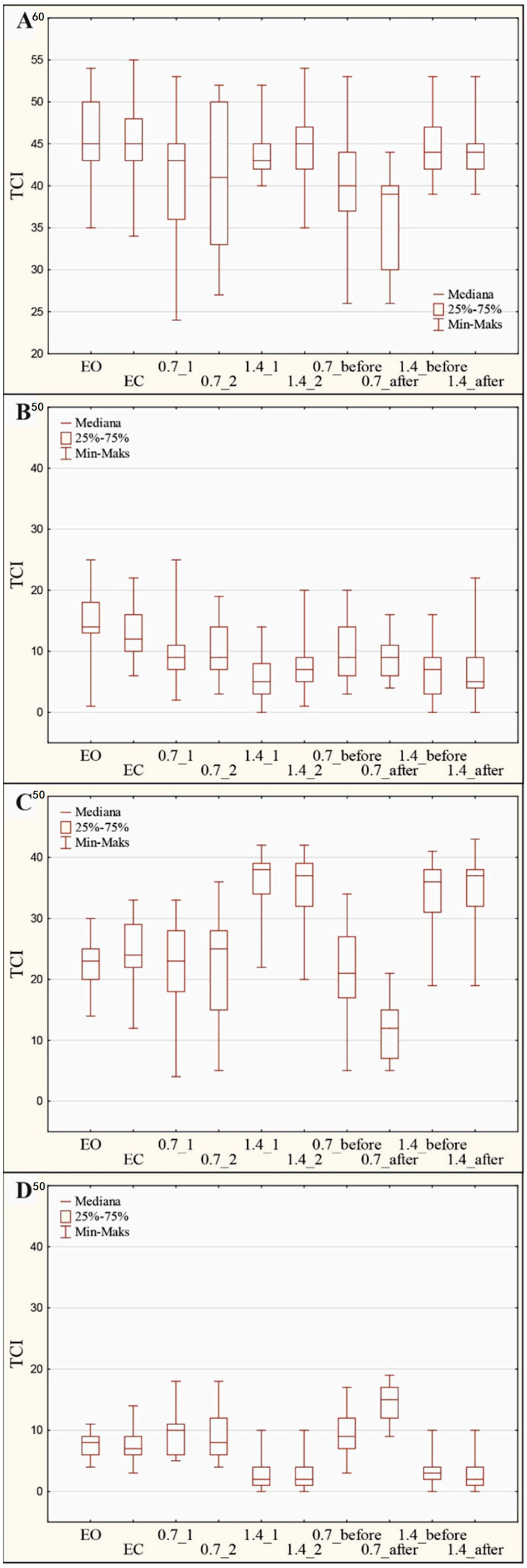

Figure 5. (**A**–**D**) TCI Values. (**A**) TCI for the whole measurement period; (**B**) TCI for the 0–0.2 s period; (**C**) TCI for the 0.2–0.5 s period; (**D**) TCI for the 0.5–1 s period.

Table 1. The p-values (p = 0.05) obtained for the comparison of TCI values. Statistically significant value differences ($p < 0.05$) (between conditions in first column and second column) are designated using the green color.

		TCI	TCI (0–0.2s)	TCI (0.2–0.5s)	TCI (0.5–1s)
OE	CE	1.00	1.00	1.00	1.00
OE	0.7_1	0.17	0.08	1.00	0.98
OE	0.7_2	0.11	0.24	1.00	0.94
OE	1.4_1	0.90	0.00	0.00	0.00
OE	1.4_2	1.00	0.00	0.00	0.02
OE	0.7_before	0.04	0.22	1.00	0.99
OE	0.7_after	0.00	0.01	0.02	0.00
OE	1.4_before	0.96	0.00	0.01	0.06
OE	1.4_after	0.94	0.00	0.00	0.01
CE	0.7_1	0.63	0.53	0.99	0.85
CE	0.7_2	0.49	0.84	0.98	0.70
CE	1.4_1	1.00	0.00	0.00	0.02
CE	1.4_2	1.00	0.01	0.01	0.08
CE	0.7_before	0.26	0.81	0.97	0.86
CE	0.7_after	0.00	0.14	0.00	0.00
CE	1.4_before	1.00	0.00	0.12	0.21
CE	1.4_after	1.00	0.00	0.03	0.05
0.7_1	0.7_2	1.00	1.00	1.00	1.00
0.7_1	1.4_1	0.97	0.03	0.00	0.00
0.7_1	1.4_2	0.34	0.91	0.00	0.00
0.7_1	0.7_before	1.00	1.00	1.00	1.00
0.7_1	0.7_after	0.14	1.00	0.06	0.07
0.7_1	1.4_before	0.92	0.09	0.00	0.00
0.7_1	1.4_after	0.94	0.40	0.00	0.00
0.7_2	1.4_1	0.92	0.01	0.00	0.00
0.7_2	1.4_2	0.23	0.65	0.00	0.00
0.7_2	0.7_before	1.00	1.00	1.00	1.00
0.7_2	0.7_after	0.22	0.98	0.07	0.14
0.7_2	1.4_before	0.83	0.02	0.00	0.00
0.7_2	1.4_after	0.87	0.15	0.00	0.00
1.4_1	1.4_2	0.98	0.68	1.00	1.00
1.4_1	0.7_before	0.74	0.01	0.00	0.00
1.4_1	0.7_after	0.00	0.19	0.00	0.00
1.4_1	1.4_before	1.00	1.00	0.98	1.00
1.4_1	1.4_after	1.00	0.99	1.00	1.00
1.4_2	0.7_before	0.09	0.68	0.00	0.00
1.4_2	0.7_after	0.00	1.00	0.00	0.00
1.4_2	1.4_before	1.00	0.90	1.00	1.00
1.4_2	1.4_after	0.99	1.00	1.00	1.00
0.7_before	0.7_after	0.44	0.98	0.09	0.07
0.7_before	1.4_before	0.60	0.03	0.00	0.00
0.7_before	1.4_after	0.65	0.17	0.00	0.00
0.7_after	1.4_before	0.00	0.41	0.00	0.00
0.7_after	1.4_after	0.00	0.85	0.00	0.00
1.4_before	1.4_after	1.00	1.00	1.00	1.00

Figure 5B shows the TCI values obtained for the period of 0–0.2 s. The introduction of disturbances into VR decreased the values of the median TCI in this period. A decrease in the median values was noticeable in the case of higher frequency oscillations in comparison to those with a lower frequency. Based on the diagram presenting the TCI values for the period of 0.2–0.5 s (Figure 5C), one can observe that higher values of the median occurred in the measurements using the scenery oscillating at a frequency of 1.4 Hz. The median values in the case of disturbances at a frequency of 0.7 Hz were similar to the OE and CE values. For the period of 0.5–1 s, the highest values were observed in the VR tests with disturbances at a frequency of 0.7 Hz (Figure 5D), whereas the lowest values were observed at a 1.4 Hz frequency.

When comparing OE and CE measurements, no statistically significant differences were observed. By contrast, statistically significant differences were observed between the results obtained in the real-life environment and those in the virtual environment for all time periods in most cases. From 0–0.2 s, statistically significant differences were observed between measurements 0.7_1 and 0.7_2, 1.4_1 and 1.4_before, and 0.7_before and 1.4_after. From 0.2–0.5 s, statistically significant differences were found between each measurement in which the oscillation frequency of 0.7 Hz was used and the measurements with the oscillation frequency of 1.4 Hz. A similar trend was observed for the 0.5–1 s period.

3.3. MACD_dS, MACD_dT, and MACD_dV

The next stage of the analysis involved the computation of the following values: MACD_dS, MACD_dT, and MACD_dV. The medians of these three values are presented in the diagrams below (Figure 6A–C).

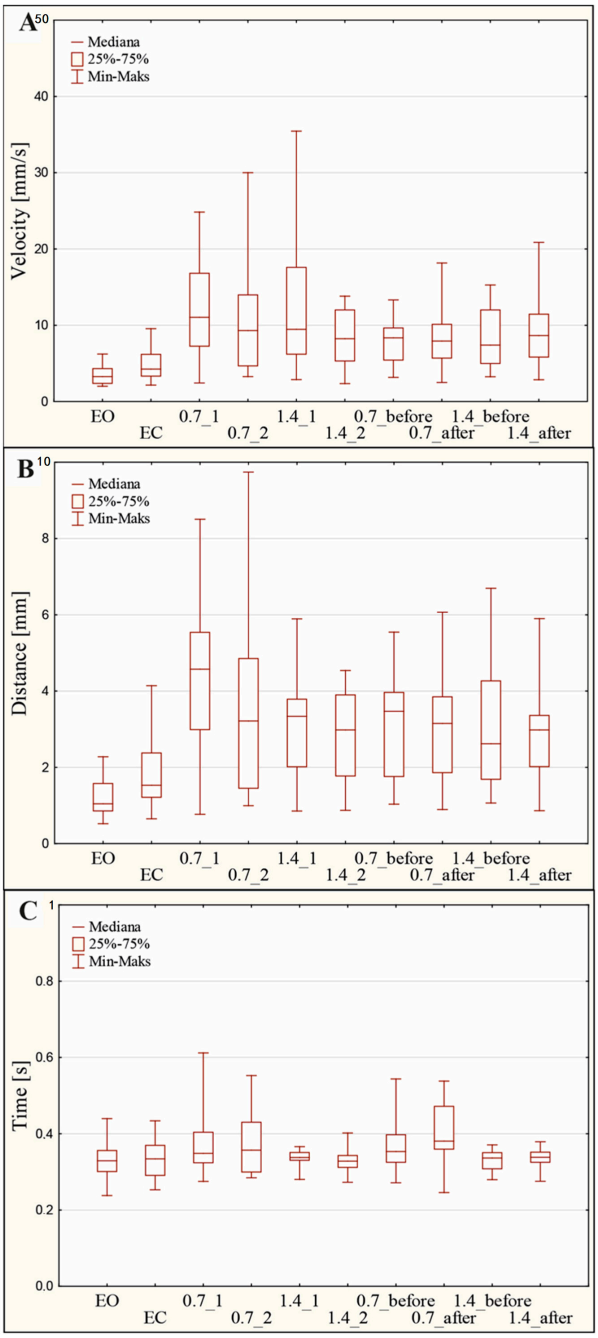

Figure 6. (**A**–**C**). Values based on the MACD algorithm. (**A**) MACD_dV; (**B**) MACD_dS; (**C**) MACD_dT.

Analyzing the MACD_dS values (Figure 6B), one can notice a significantly increase in the obtained values in the VR tests in comparison with the measurements taken in the real-life environment—both in the median values and the interquartile distribution of the data. No statistically significant differences were found when comparing the OE and CE tests. For the real-life environment and VR results, statistically significant differences were observed between the OE tests and all measurements in the virtual environment as well as between the CE tests and the 0.7_1, 0.7_2, 1.4_1, 0.7_before, 0.7_after, and 1.4_after measurements. When comparing the measurements obtained using VR between individual tests, no statistically significant differences were found.

In the case of the analysis of the MACD_dT values (Figure 6C), significantly higher values were obtained for the measurements conducted at a frequency of 0.7 Hz. Statistically significant differences were noted while comparing the measurements in the real-life environment (OE and CE) and the 0.7_after, 1.4_before, and 1.4_after measurements.

Figure 6A presents the diagram of the median MACD_dV values for individual measurements. Importantly, the closure of eyes increased the MACD_dV values. A similar trend took place in the case of the introduction of VR; the MACD_dV values were higher in the measurements with the oscillating scenery. Also, an increase in the data distribution was observed in the 0.7_after and 1.4_after tests in relation to measurements 0.7_before and 1.4_before. The analysis of the MACD_dV values did not reveal any statistically significant differences in the OE and CE tests. When comparing the values obtained in the real-life environment and the virtual environment, statistically significant differences were observed between OE and all measurements using the virtual scenery, as well as between CE and the 1.4_1, 0.7_1, 0.7_2, and 1.4_after measurements. No statistically significant differences were found when comparing the measurements in VR.

4. Discussion

4.1. Data Analysis in the Time Domain

The analysis of stabilographic values, such as the COP velocity and the COP movement range, constitutes the basis for evaluating a human being's ability to keep balance [1,5,16]. The introduction of oscillation into the virtual scenery in the AP direction led to a statistically significant increase in values in both the cases of the COP velocity and the COP movement range in the AP direction in relation to the measurements with OE (Figure 4A,B). Similar conclusions were drawn in previous research conducted by the authors [1]. The increase in these values results from the balance of the body following the oscillating scenery, which was confirmed by the authors' preceding investigations, where FFT analysis proved the presence of an additional harmonic at a frequency equal to the frequency of the scenery movement. The investigations also demonstrated the presence of additional harmonic at other frequencies, which may indicate a change in the balance-keeping strategy or destabilization of the tested person [20,21].

In the measurements with the change in frequency (0.7/1.4 and 1.4/07), one could observe a greater interquartile distribution for the data obtained after the frequency change (0.7_after and 1.4_after) in the case of the COP velocity values in the AP direction in comparison with the data obtained before the frequency change (0.7_before and 1.4_before) (Figure 4A,B). This result may mean that the change from one frequency to another exerts a destabilizing impact and somehow forces the tested participant to search for a new strategy to maintain balance.

Analyses in the time domain and additional analyses in the frequency domain make it possible to determine if and how virtual disturbances influence changes in the balance-keeping strategy [21] and body-balancing strategy during movement [22]. However, the frequency analysis is limited to the discovery of movement components of a cyclic character. In addition, such an analysis is burdened with noise related to spectral leakage and incomplete signal periods in the measurement window. The application of computational methods drawn from stock exchange analyses makes it possible to supplement the performed analyses with signal components that do not have a cyclic character [13,14].

4.2. Trend Change Index (TCI)

An increase in the mean value of the COP velocity or movement range may indicate problems with maintaining balance. There is evidence, based on the analyses of COP movement in the frequency domain, that the lines appearing in FFT diagrams may be used for the assessment of the balance-keeping strategy. For instance, the movement of the body following the cyclically oscillating virtual scenery, which may be observed in the FFT or short-time Fourier transform (STFT) diagrams in the form of lines, can be used for this purpose [1]. Studies performed by Bizid R. et al. [23] and Micarelli A. et al. [24] indicate that there might be a connection between the frequency of COP oscillation and the strategy of maintaining balance. They claim that low frequencies (0–0.5 Hz) mostly account for visual–vestibular regulation, medium frequencies (0.5–2 Hz) for cerebellar regulation, and high frequencies (>2 Hz) for proprioceptive regulation.

However, the above-described frequency analyses enable only the detection of the signals of a cyclic character, whereas the real COP movement may include both cyclic and non-cyclic components [16]. The developed TCI, calculated using the detection of trend changes based on stock exchange analysis methods, makes it possible to extend the analyses and supplement them with a non-cyclic component of the signal [19,25]. The TCI value shows how often the COP was subjected to trend changes during the movement performed during the whole measurement (Figure 4A,B).

The conducted investigations demonstrated that in most cases, the value of the TCI, which was calculated for the whole interval from 0 to 1 s, did not change significantly, irrespective of the measurement conditions (Table 1). Such constancy, independent of the measurement conditions, may signify that to maintain a stable body posture, even in a situation of visual stimulation, a certain number of trend changes is required, which enables the correct functioning of the proprioceptive and visual–vestibular systems [13,16]. An important exception occurs in the case of the measurement of the scenery oscillating at a frequency of 0.7 Hz, which took place after preceding oscillations at a frequency of 1.4 Hz. There was a clear decrease in the TCI values in relation to other measurements, i.e., the number of discovered changes in the trend, which may indicate a calmer movement at a frequency of 0.7 Hz when it occurs after a frequency of 1.4 Hz. Additional interpretation of this fact is supported by the TCI values determined individually for three time periods equal to the time between subsequent trend changes. The diagrams showed a significant decrease in the TCI values in the 0.2–0.5 s period and an increase in the interval of 0.5–1 s (Figure 5A–D), denoting considerable calming of COP movement. There appeared to be fewer 'leaps' of the COP lasting shorter than 0.5 s in favor of a greater number of leaps lasting longer than 0.5 s. This interval included COP movements performed at a frequency of 0.7 Hz. Therefore, from a statistical perspective, the tested individuals were more prone to disturbances at a frequency of 0.7 Hz if they had been previously exposed to disturbances at a higher frequency. Notably, such differences were undetectable based on the analysis of the COP mean velocity or the COP movement range, where a statistically significant difference appeared only between the OE and 0.7_after tests but did not occur in other cases (Figure 4A,B).

The analysis of measurements conducted at a frequency of 1.4 Hz revealed a significant increase in the TCI values in relation to other measurements in the 0.2–0.5 s period and a decrease in the 0.5–1 s period. The increase in the 0.2–0.5 s period was related to the fact that the body follows the moving scenery. This interval includes the 1.4 Hz frequency, which is also visible in the analyses in the frequency domain. This difference was noticeable, particularly in relation to the OE and CE measurements. In the remaining two time periods, 0–0.2 s and 0.5–1 s, at a frequency of 1.4 Hz, there was a decrease in the TCI values in relation to the measurements with OE and CE.

COP movement is connected not only to the constant loss of balance and the return to the balance-keeping position but also with the fact that it constitutes a source of information for the balance-keeping systems on what is happening with the body. The lack of significant changes in the total TCI value in individual measurements (Figure 4A), i.e., the number of trend changes in COP movement, leads to at least two interpretation possibilities. First, the lack of a considerable increase in the number of trend changes in COP movement by a given person, i.e., an increase in the number of trend changes in the interval corresponding to 1.4 Hz, which occurs at such disturbances of the scenery, forces an increase in these changes in other time periods. The second interpretation may be that a healthy person does not need to maintain a constant number of trend changes in each of the indicated three time periods. What is essential in the case of these individuals is the total number of such changes. Both hypotheses may be particularly important in the case of testing participants with balance-keeping deficits. However, in the case of such individuals, further studies and investigations regarding these hypotheses are required to be scientifically useful.

4.3. MACD_dS, MACD_dT, and MACD_dV

The determined MACD_dS, MACD_dT, and MACD_dV values (Figure 6A–C) make it possible to further extend the analyses performed based on the trend change detection using the methods of stock exchange analysis [21]. The obtained values of the MACD_dV strongly correlate with the values of the COP mean velocity, which seems obvious. However, the knowledge of the time between trend changes and the distance covered by the COP at that time enables the collection of additional information on COP movement to be obtained [18]. Information on the changes in the MACD_dT and MACD_dS values may indicate if the resulting changes in the velocity values, or their lack thereof, result from the change in the length of individual leaps of the COP, the time in which such a leap takes place, or from both of these values simultaneously.

Statistically significant differences between the OE and CE measurements and all other measurements occurred only in the case of the MACD_dS, where an increase in the median value was observed (only measurements between OE and 1.4_2 lacked significance). At the same time, most cases displayed a lack of statistically significant differences at the MACD_dT time. This showed that, in these cases, the observed increase in the velocity values occurred because of an increase in the mean distance between the subsequent COP locations in which the trend change took place, whereas the mean time between those leaps remained constant. An exception was the measurement performed for 0.7_after, where statistically significant differences were noted in the OE and CE measurements. A simultaneous increase in the MACD_dS and MACD_dT values caused a lack of change in the velocity between 0.7_after and CE, which could be seen both in the case of MACD_dV and the COP mean velocity.

What is worth mentioning is the fact that no case showed a statistically significant decrease in the MACD_dT value. A MACD_dT value that does not significantly decrease is essential from the perspective of the ability to maintain balance by the tested individuals. A simultaneous increase in MACD_dS and a decrease in MACD_dT might indicate much longer leaps of the COP performed at a shorter time. Such a case might potentially lead to destabilization and a subsequent fall of the tested person [25,26].

5. Conclusions

The use of stock exchange indices to assess human body stability complements standard analyses in both the time and frequency domains. Analyzing TCI, MACD_dV, MACD_dT, and MACD_dS values provides additional insight into factors affecting standard parameters like path length, mean velocity, and movement range. By integrating trend change analysis with stabilographic parameter analysis, we can glean information on posture correction frequency, intervals between corrections, and COP movement speed. Unlike FFT analysis, our time domain-based algorithm ensures the results are unaffected by noise. Decomposing the TCI analysis into intervals reveals shifts in COP movement patterns, indicating changes in balance strategy.

TCI, MACD_dT, and MACD_dS analyses could prove valuable in testing patients with balance disorders. Elevated MACD_dS values coupled with shortened MACD_dT may signify increased fall risk. Deviations in total TCI value and intervals compared to healthy individuals may indicate balance-related dysfunctions, necessitating further investigation in specific patient groups.

However, our algorithm's sensitivity is limited to rapid postural corrections due to short time windows and MACD parameter values. Future plans involve consulting neurologists to refine the methodology and application, particularly in diagnosing neurological diseases.

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PLOS ONE

RESEARCH ARTICLE

Trend change analysis in the assessment of body balance during posture adjustment in reaction to anterior-posterior ground perturbation

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Abstract

Postural adjustments (PA) occur to counteract predictable perturbations and can be impaired as a result of musculoskeletal and neurological dysfunctions. The most common way to detect PA is through electromyography measurements or center of pressure (COP) position measurements, where analysis in time domain and frequency domain are the most common. Aim of the research was to determine whether a new method of analyzing stabilographic measurements—the COP trend change analysis (TCI) of temporary posture corrections- can expand understanding of changes in balance strategy connected with PA. The study group involved 38 individuals (27women, 11men) aged 23±2.6 years. Measurements were performed using a stabilographic platform placed on a perturbation platform. The tests involved three measurements with forward and backward momentary movements of the platform. Participants were tested in three conditions–knowing the nature, time and direction of perturbation (Tr3), knowing only the nature of perturbation (Tr2) and without any information about the perturbation (Tr1). Statistically significant differences were revealed in the last second of Tr3 for the mean velocity of COP ($p<0.05$) and for two TCI parameters–TCI_dV ($p<0.05$) and TCI_dS ($p<0.01$). The increase in TCI_dV was related to the increase in the mean distance between trend changes (TCI_dS) and constant value of the mean time between trend changes (TCI_dT). The increase of the mean value of TCI_dS was the result of smaller number of posture corrections with the distance 0–2 mm and lager number with the distance 4–6 mm. Obtained results proved that the TCI analysis is a method enabling an extended analysis of PA, indicating the nature of changes occurring in posture corrections–longer momentary jumps of COP–related to a change in the strategy of maintaining balance before a known disorder, which has not been analyzed in this type of research so far.

Competing interests: The authors have declared that no competing interests exist.

Abbreviations: Tr1, test no. 1 where the test participant did not know the nature, starting time, or direction of perturbation; Tr2, test no. 2, where the test participant knew the nature of perturbation but did not know its starting time and direction; Tr3, test no. 3, where the test participant was informed (using a countdown display) about the starting time and direction of perturbation; -6 s . . .. -1 s, period of time preceding perturbation: -6th second preceding perturbation (-6 s represents the time between the 6th and the 5th second before perturbation) to the moment of perturbation (-1 s represents the time between the 1 second preceding perturbation and the perturbation itself); AP, anterior-posterior; ML, mediolateral; FFT, fast Fourier transform; COP, center of foot pressure; COM, center of mass; APA, anticipatory postural adjustment(s); EPA, early postural adjustment(s); Fpass, filter pass frequency; Fstop, filter cut-off frequency; TCI, total number of trend changes during the entire test; TCI_dT, mean value of time between successive trend changes; TCI_dS, mean value of displacement between successive trend changes; TCI_dV, mean value of the COP displacement velocity between successive trend changes; Vcop, resultant COP displacement velocity; IMU, inertial measurement unit; EMG, electromyography.

Introduction

Posture is not just about the simple, static positioning of body parts. It is a dynamic process that involves constant modifications to ensure balance while carrying out different activities. Postural adjustment (PA) and compensation are natural reactions of the body to destabilizing stimuli [1–3]. PAs can be early (EPAs) or anticipatory (APAs). Research conducted by Xie *et al.* [4,5], revealed that PAs may result from the internal initiation of movement (preparation to perform a certain activity). Similarly, Cleworth *et al.* [1] and Sibley and Etnier [6] indicated that a destabilizing stimulus may come from the environment. This conclusion was confirmed by Mohapatra and Krishnan [3], who also indicated correlations between the intensity of the APA and EPA and the intensity of the environmental stimulus effects. The tests performed by Ritzmann et al. [7] demonstrated the possibility of training PA resulting in faster post-perturbation balance recovery. Previous authors' research [8] demonstrated the correlation between the intensity of the PA and the post-perturbation shift of the COP, which indicated that the PA mattered in the process of balance recovery.

PA can be observed by measuring changes in muscle activity, what is the most common method, but measurements of changes in the center of mass (COM) or the center of pressure (COP) displacements can also be used. In the case of COP measurements, PA is visible in the increase in COP velocity [9,10]. Other research revealed that the ranges of the COP and COM displacements during PA and compensation are directly correlated with the likelihood of falling [4,11,12].

Research conducted by Vuillerme and Nafati [13] revealed a change in the median of the main frequency of the COP movement during the concentration and preparation for an induced voice stimulus. The frequency of the COP movements can be analyzed when searching for cyclic changes using fast Fourier transform (FFT)-based algorithms [13–15], and wavelet analyses [16,17]. A new advancement in postural analysis is the application of trend change analysis (TCA). This method can identify rapid adjustments and an elongation in the displacement between consecutive postural adjustments [18,19]. Borrowed from strategies initially used in stock market analyses, TCA aids in measuring postural adjustments in both the front-to-back (A/P) and side-to-side (M/L) directions. Furthermore, it enables the computation of the count of adaptations and the time gap between successive posture adjustments, offering insights into how the body reacts to postural difficulties [20–22]. Such analysis may explain the origin of changes observed in other quantities, such as velocity as well as take into account non periodic postural correlations, which are not included in frequency analysis.

Taking the above into account, we hypothesize that the observed increase in velocity of COP is connected with the change of balance strategy, what should be visible in the changes in trend changes parameters in the COP signal, such as number of posture corrections as well as time and distance between following posture corrections. The aim of the study was to check if different conditions of PA examination would result in differences in values of trend changes parameters. Positive verification of the hypothesis may allow the formulation of a new method supporting assessment of PA. The reliability and validity of trend changes parameters was confirmed in previous research [18,19].

The tests presented in this article are the continuation of the research aimed at detecting the PA and accompanying phenomena related to COP movements, when preparing for postural perturbations connected with the abrupt movement of the ground [8]. These measurements made it possible to assess muscle activation in EPA and APA time intervals before movement of the basis.

Materials and methods

Study group

The study group consisted of 38 persons (27 women, 11 men) aged 23 ± 2.6 years, height 172 ± 9.6 cm and weight of 70 ± 17 kg. Excluding factors: past injuries of lower limbs and balance problems—both confirmed by physiotherapists and declared by potential participants. Each of the study participants gave written informed consent to participate in the study. This study had been previously approved by the Ethics in Research Committee of the Academy of Physical Education in Katowice (protocol number 5/2020).

Experimental procedure

The measurement stand was composed of a platform for the COP distribution measurements (WinFDM-S, Zebris, 100 Hz data acquisition frequency, 2560 tensiometer sensors, 34 cm x 54 cm sensors area) fixed to the central part of the perturbation platform in the form of the treadmill for the training and prevention of postural perturbations (BalanceTutor, MediTouch) (Fig 1). Pre-test preparations involved the attachment of the EMG electrodes (Ultium EMG, Noraxon, 2000 Hz data acquisition frequency) and inertial measurement unit (IMU) sensors (Ultium Motion, Noraxon, 200 Hz data acquisition frequency) [8]. Electrodes were attached to musculus tibialis anterior, musculus rectus femoris, musculus gastrocnemius medialis and musculus gastrocnemius lateralis—muscles actively involved in the process of maintaining balance in AP direction (a detailed description of the procedure for analyzing the recorded EMG signals along with the results can be found in the supplementary materials: S5 File). One IMU sensor was attached to the treadmill belt enabling the detection of platform motion initiation and synchronization of all devices.

A participant stood barefoot on a stabilographic platform protected against falling. Each study participant was protected against falling using a special harness attached above the balance platform (Fig 1). This device neither restricted movements nor affected their freedom. The participant's task was quiet standing with arms along the sides during each of the measurements (Tr1,Tr2 and Tr3).

During individual measurements:

- Tr1 –participant did not know the nature, starting time, or direction of perturbations,
- Tr2 –participant knew the nature of perturbation but did not know its starting time and direction—participant was informed that the perturbation would be of the same nature as in the Tr1 trial,
- Tr3 –participant was informed (using a display) about the starting time and the direction of perturbation. The countdown was auditory and was also displayed 5 s before perturbation. The countdown intervals amounted to 1 s.

The same participants were involved in the three measurements (Tr1,Tr2 and Tr3). Each measurement lasted 30 seconds and consisted of two movements of the perturbation platform: the first one–forward–that took place 10 s after the initiation of measurements and the second one–backward– 20 s after the initiation of measurements. Both movements were identical, with a length of 9,5 cm and a duration of 0.5 s. The test was repeated three times. In case of balance loss the measurement was repeated.

Processing of results

After taking the measurements the results were processed as follows:

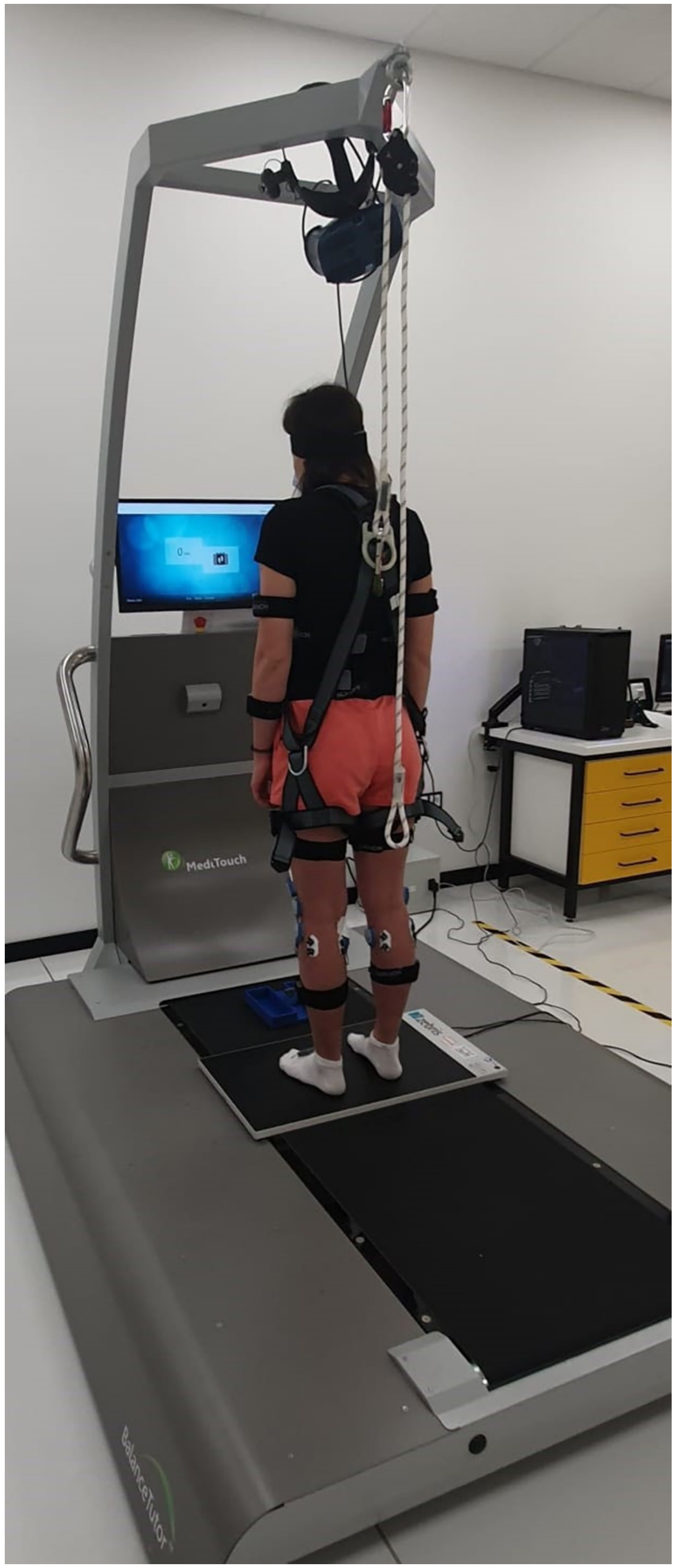

Fig 1. Measurement stand and a test participant.

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i. determination of perturbation platform movement initiation using the sensor attached to the treadmill belt—detection of the beginning of movement based on the detection of acceleration increase,

ii. determination of the COP displacements in the AP and ML directions—COP waveform exported from the measurement platform,

iii. signal filtration using a low-pass Butterworth IIR filter (Fpass = 10 Hz, Fstop = 12 Hz, Astop = -60 dB),

iv. division into sections lasting 1 s each, counting from the 6th second preceding perturbation to the moment of platform perturbation.

For each 1-second section, the average velocity of the COP displacement was calculated (Vcop), and an analysis of trend changes was performed.

Trend changes index calculation

The method of trend change analysis comes from technical analysis of signals, which is based on moving averages with exponential weights. To begin with, the algorithm employs the MACD (Moving Average Convergence Divergence) indicator calculation process in its initial phase [18,19]. This algorithm assesses relationships linked to the convergence and divergence of moving averages of a measured signal—in the study it is COP signal. In the initial calculation, it was determined the MACD line for the COP signal, by computing the difference between two Exponential Moving Averages (EMAs) with lengths of 12 and 26 samples, as per Eqs 1 and 2.

$$MACD = EMA_{COP,12} - EMA_{COP,26}. \qquad \text{Eq1}$$

where:

EMACOP,12—faster exponential moving average for COP signal,

EMACOP,26—slower exponential moving average for COP signal.

$$EMA = \frac{p_0 + (1-\alpha)p_1 + (1-\alpha)^2 p_2 + \cdots + (1-\alpha)^N p_N}{1 + (1-\alpha) + (1-\alpha)^2 + \cdots + (1-\alpha)^N}. \qquad \text{Eq2}$$

where:

p0 –ultimate value,

p_1 –penultimate value,

p_N–value preceding N periods,

N = number of periods,

α = a smoothing coefficient equal to 2/(N+ 1).

Moving on to the subsequent phase, the signal line is calculated as an EMA with a length of 9 samples from the MACD line signal in accordance with Eq 3.

$$Signal\ line = EMA_{MACD\ line,9}. \qquad \text{Eq3}$$

The points where the MACD line intersects with the Signal line are pivotal in identifying shifts in the trend of the COP signal (Fig 2). These intersections, in terms of quantity, determine what it refers to as the Trend Changes Index (TCI).

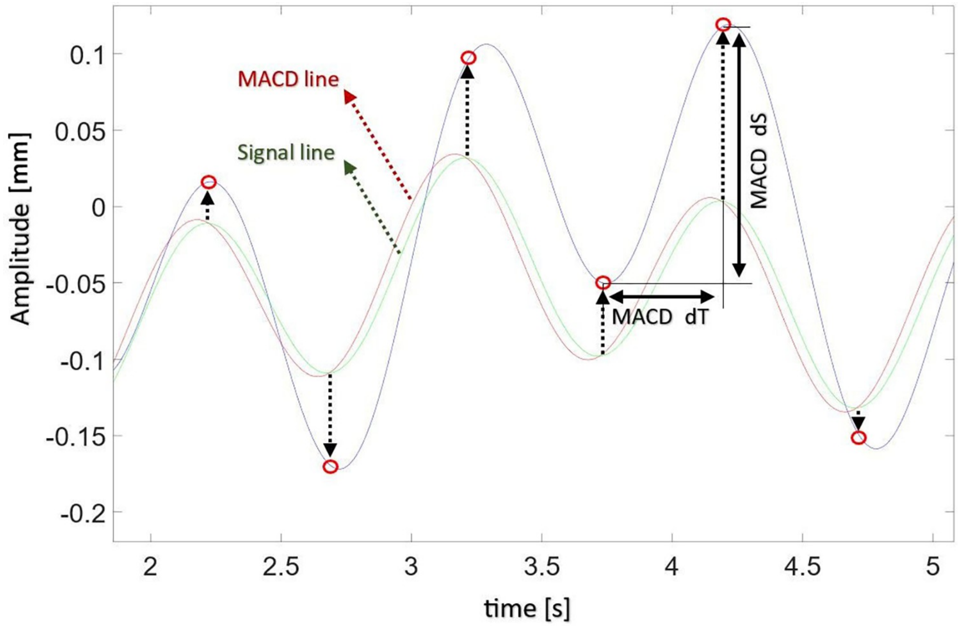

Fig 2. Graphical interpretation of determined parameters and places of trend changes calculated on basis of MACD algorithm. The blue line shows the signal course, the red circles mark the places of trend changes detected by the algorithm. The Signal line is marked in green and the MACD line is marked in red.

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Basic trend changes analyses

Moving forward, it was proceeded to calculate the time sections between successive trend change points in the COP signal (Fig 2). This process led to the creation of the MACD_dT table (table contains of detected MACD_dT values), whose average value for the entire analyzed course is denoted as TCI_dT. Consequently, we calculated the displacements between consecutive trend change points, resulting in the MACD_dS table. The average value of this table corresponds to TCI_dS. Lastly, the MACD_dV table is established by dividing the values in the MACD_dS table by their corresponding MACD_dT table. The TCI_dV indicator is then determined as the average value of the MACD_dV table.

The following quantities were determined separately for AP and ML directions:

- TCI–total number of trend changes during the entire test,
- TCI_dT–mean value of time between successive trend changes,
- TCI_dS–mean value of displacement between successive trend changes,
- TCI_dV–mean velocity of the COP displacement between successive trend changes.

Resultant values were calculated as follows:

- for TCI, results obtained for both directions were summed together,
- for other quantities, resultant value determined on the basis of the AP and ML components as the square root of the sum of squares of the values calculated in each direction.

The forward and backward displacements of the platform are analyzed as one group because the movement of the COP before the displacement is analyzed. Changes of feature of the COP signals are important in these analyzes and they can be manifested in the number of

Table 1. TCI values in successive moments before perturbation in tests Tr1, Tr2, and Tr3. The results concern the measurements from the 6th second preceding perturbation (-6 s represents the time between the 6th and the 5th second before perturbation) to the moment of perturbation (-1 s represents the time between the 1 second preceding perturbation and the perturbation itself).

Test	*Tr1*						*Tr2*						*Tr3*					
Time	*-6 s*	*-5 s*	*-4 s*	*-3 s*	*-2 s*	*-1 s*	*-6 s*	*-5 s*	*-4 s*	*-3 s*	*-2 s*	*-1 s*	*-6 s*	*-5 s*	*-4 s*	*-3 s*	*-2 s*	*-1 s*
Median	*7*	*8*	*8*	*8*	*8*	*8*	*8*	*8*	*8*	*8*	*8*	*8*	*7*	*7*	*8*	*8*	*8*	*7*
Quartile 1 (25%)	*5*	*6*	*6*	*6*	*6*	*6*	*5.8*	*6*	*6*	*6*	*6*	*6*	*5*	*6*	*6*	*6*	*6*	*6*
Quartile 3 (75%)	*9*	*10*	*10*	*10*	*10*	*10*	*10*	*10*	*10*	*10*	*10*	*10*	*9*	*9*	*10*	*9*	*10*	*9*
Mean	*7.53*	*8.13*	*8.04*	*8.06*	*8.60*	*8.32*	*8.03*	*8.26*	*8.16*	*8.33*	*8.31*	*8.57*	*7.36*	*7.50*	*8.08*	*7.92*	*8.05*	*7.77*
SD	*3.12*	*3.05*	*2.71*	*3.19*	*3.10*	*2.82*	*3.17*	*2.98*	*3.09*	*2.93*	*3.04*	*2.94*	*2.74*	*2.74*	*2.55*	*2.96*	*3.22*	*2.65*

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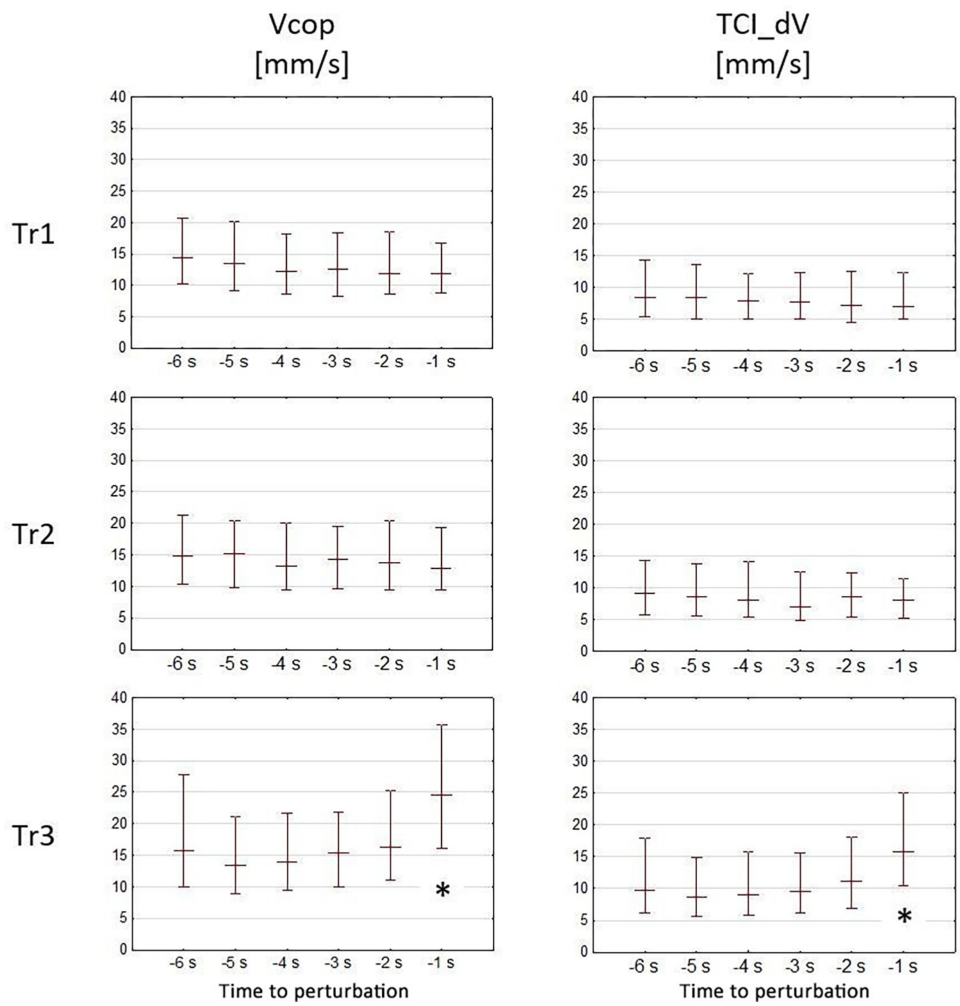

Fig 3. Values of Vcop and TCI_dV in relation to moments preceding perturbation in tests Tr1, Tr2, and Tr3. The chart presents the medians and the interquartile range for measurements between the 6th second before perturbation (-6 s represents the time between the 6th and the 5th second preceding perturbation) to the moment of perturbation (-1 s represents the time between the 1 second before perturbation and the moment of perturbation). Statistically significant values differences ($p < 0.05$) are designated using "*".

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trend changes of the COP signals—these changes are related to the number of posture corrections, regardless of the direction in which this correction occurs.

Histogram analysis

The more precise indication how increase in TCI_dS in the last second before the disturbance in the Tr3 test was corelated with changes in MACD_dS values required division into smaller displacement intervals of this quantity. This division was presented on histograms containing the number of individual MACD_dS's (expressed as a percentage) in each displacement interval divided into subsequent time sections before the disturbance. The interval discretization step in the TCI_dS displacement histogram charts amounted to 2 mm, whereas results were presented up to a maximum displacement of 30 mm. The tests did not reveal TCI_dS values exceeding 30 mm. For each subject and each test, the TCI_dS histogram chart was calculated before perturbation. The diagrams present the median values obtained for the group of test participants. The values of changes (*i.e.*, number of displacements) per each displacement interval are expressed as a percentage. The sum of all of the displacements in relation to each displacement interval subjected to analysis was adopted as 100%.

Statistical analysis

The Shapiro-Wilk test did not reveal the presence of normal distributions in relation to all parameters, therefore, median values were used in all analyses. The Friedman test, followed by the Pairwise Wilcoxon with a post-hoc Holm correction test, were used to compare the parameters between the Tr1, Tr2, and Tr3 tests. In addition, the size of the effect was calculated using the methodology proposed by Rea and Parker [23]. The difference between the medians was reported as statistically significant as long as the effect was high (0.8 and more) or at least medium (0.5–0.8). Small effects were not reported.

Results

Analyzes of the EMG signals obtained for the described tests, which were presented in detail in the article [9], showed differences in the activation of the lower limb muscles between trials Tr1, Tr2 and trial Tr3. The information that was used from the analysis was mainly: the presence of PA (in the form of APA and EPA) was objectively confirmed in the Tr3 trial and the time at which PA occurred in the EMG signal was determined, which, as it turned out, did not exceed one second before the disturbance. It was shown that in the Tr3 trial, EPA and then APA appear during the one second preceding the disturbance (S2 and S4 Files). This made it possible to indicate the time intervals of both these reactions–EPA and APA. It was assumed, that different muscle activations bring about different characteristics of COP signals in the detected time intervals, what should be visible in different values of analyzed quantities. Taking into account this assumption, analyzes of the COP movement were carried out for 6 seconds preceding the introduced perturbation assuming that PAs may be detected in the last seconds.

Results from the -6th to the -1st second (Table 1, S1 File) are identical (the TCI median is always an integer for an odd number of measurements). The TCI medians do not differ significantly between successive time sections in the Tr1, Tr2, and Tr3 tests.

In the case of TCI_dT, there were no statistically significant differences in the values between successive moments in all of the tests. However, in the case of TCI_dS (similar to TCI_dV and Vcop), it was possible to notice a significant increase in the distance ($p < 0.05$) in the Tr3 test at -1 s in relation to the measurements between -6 s and -2 s. A graphical presentations of the Vcop and TCI_dV values are presented in Fig 3, while the TCI_dt and TCI_ds values are presented in Fig 4.

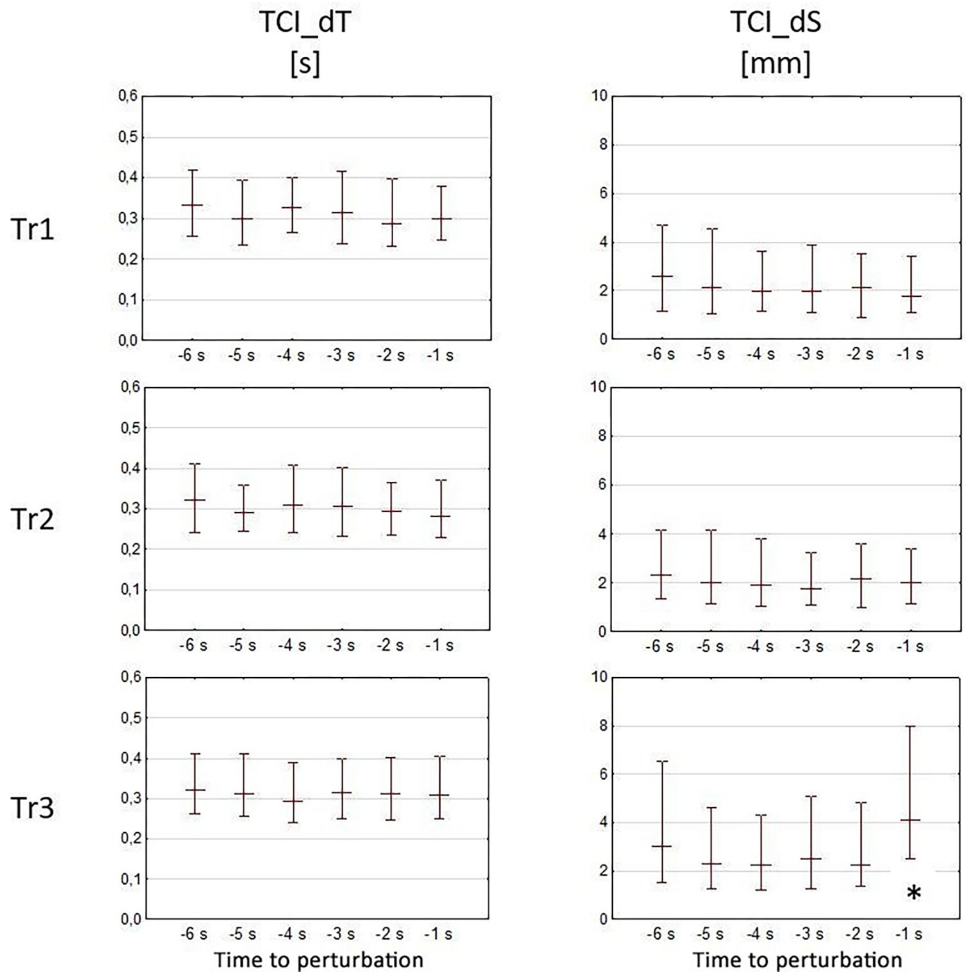

Fig 4. Values of TCI_dT and TCI_dS in relation to successive moments preceding perturbation in tests Tr1, Tr2, and Tr3. The chart presents the medians and the interquartile range for the measurements between the 6th second before perturbation (-6 s represents the time between the 6th and the 5th second preceding perturbation) to the moment of perturbation (-1 s represents the time between the 1 second before perturbation and the moment of perturbation). Statistically significant differences ($p < 0.05$) are designated using "*".

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An increase in the TCI_dS values in the Tr3 test at -1 s significantly affected the change in TCI_dV. At the same time, an increase in the TCI_dS values also indicated a change in the balance-maintaining strategy at the aforesaid second of the movement. To further investigate the change of the strategy (in accordance with the methodology), histogram charts of TCI_dS at successive time sections preceding perturbation were plotted (Fig 5).

The number of displacements expressed as a percentage (median values, Fig 5) was subjected to the comparative analysis for each displacement interval. In all comparisons the ANOVA test was used to compare the medians of 15 groups of displacement interval in relation to all time sections and in accordance with the direction of the arrow "a" (Fig 5 Tr3 b) from the range of 0–2 mm to 28–30 mm (S3 and S6 Files). In each of the subgroups (where each subgroup contains values of number of displacements for all time intervals), it was possible to obtain several post-hoc comparisons. Statistically significant differences ($\alpha = 0.05$) were obtained between 0–2 mm and 4–6 mm sections, indicated in Fig 5 Tr3 A) as arrow "I" and arrow "II". The number of displacements in the displacement section 0–2 mm at -1 s time

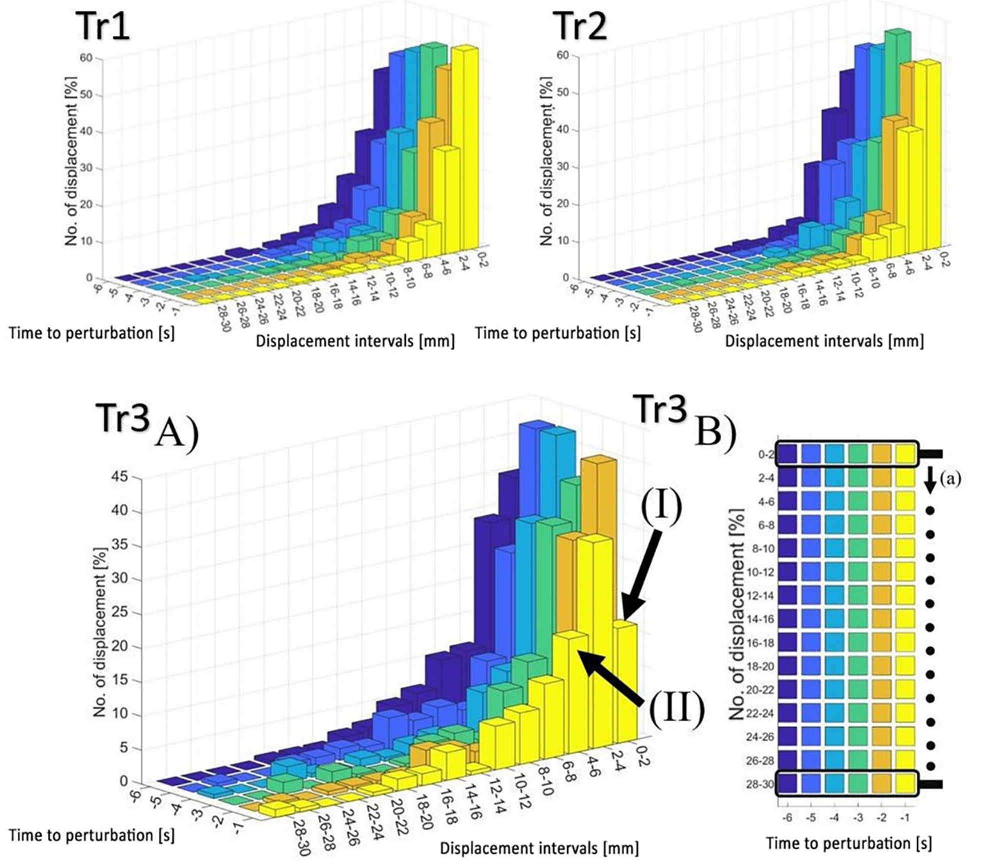

Fig 5. Histogram charts of TCI_dS in relation to Tr1, Tr2 and Tr3 trials. The number of displacements is expressed as a percentage. The sum of all of the displacements in relation to each displacement interval subjected to analysis was adopted as 100%. Tr3 A) Histogram charts of TCI_dS in relation to Tr3 at successive time sections preceding perturbation. The arrows I and II represent places of statistically significant changes in values between the time -1s and earlier time sections. Tr3 B) Top view of the TCI_dS histogram chart where the boxed-in areas designate the tests in relation to which the analysis involving the use of statistical methods was performed. The arrow (a) presents the sequence of analyses in related groups at increasing displacement intervals.

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section (arrow I) in relation to the previous time periods (-6 s to -2 s) decreased ($p<0.01$). Conversely, the number of displacements in the displacement section 4–6 mm at -1 s (arrow II) in relation to previous time periods increased ($p<0.01$).

Discussion

Muscle activity analysis is the basic method used to detect and analyze PA phenomena [8,22]. COP observation is used much more seldomly. Research conducted by Slijper *et al*. [24] and Krishnan *et al*. [25] are two examples that used COP analysis in PA detection. Both authors noticed an increase in the support path length in tests with suspected PA phenomenon. Bouisset [10] and Chander *et al*. [9] in their research indicated an increase in Vcop in relation to the baseline measurements. They also discovered that Vcop may be a better index to assess the PA phenomenon than COP displacements in AP and ML directions. All these observations were confirmed in described research.

Previous analyses [8] of the EMG signals obtained for these measurements proved the presence of EPA and APA in the Tr3 trial during the last second preceding the disturbance.

That was confirmed in current analyses of COP movement, where a statistically significant increase in Vcop was noticed in the last second preceding perturbation in Tr3 (Fig 3) which was not observed in tests Tr1 and Tr2. The similar increase can be observed in TCI_dV calculated from TCI_dT and TCI_dS what can confirm correctness of the assumptions of the developed method of trend changes analysis.

A surprising fact was the impossibility to observe changes in TCI numbers in subsequent time sections, particularly in test Tr3 (*i.e.*, where the test participant knew when the moment of unbalance would take place). It would seem that increased Vcop and TCI_dV values should result in a different TCI number as a reaction of the nervous system aimed to coordinate the activity of individual muscles [26–29]. Additional analysis of TCI_dS and TCI_dT revealed that the observed increase in both velocities was primarily due to the extension of the path of momentary leaps of COP (TCI_dS) while maintaining constant times between leaps (TCI_dT) and consequently the same number of TCI.

What interesting opposite changes in balance strategy were observed in research comparing results obtained for people with Parkinson Disease and healthy participants during normal standing [18]. Measurements revealed higher values of Vcop and TCI_dV, just like in measurements described in this article, but smaller values of TCI_dT in PD group with no statistically significant differences in TCI_dS. This may indicate that there are at least two different mechanisms of adjusting the balance to destabilizing factors, possible to detect by means of trend change analysis–a physical disorder and, in mentioned research, Parkinson Disease.

Taking into account the fact that COP is a reaction to COM movement, registered, in this article, changes in TCI_dS could result from the change in the COM movement. However, taking into consideration the fact that the COP movements are identified as a resultant force acting the ground [30,31], it could also indicate the changed patterns of the work of muscles preparing for oncoming disruption, which is often observed in such types of tests [8,24,32,33]. However, the statement requires additional research.It should be noted that the lack of change in the number of trend changes could also result from the very short measurement time precluding the detection of slight differences.

The balance-maintaining strategy based on the TCI_dS charts

Temporary posture corrections are one of the crucial elements related to the balance strategy. The developed TCI coefficient allows for a clear indication of the number of posture corrections. It can be expected that this number should be correlated with the average COP velocity, which, how it was mentioned above, was not observed in the conducted research. Despite the increase in Vcop in the last second before the perturbation in trial Tr3, the number of TCI in this interval remained unchanged. Observed differences in TCI_dS and lack of such changes in TCI_dT may affect the change in COP velocity, without the need to change the total number of trend changes (TCI). The research proved however that despite the constant value of total TCI, there are changes in number of COP displacements, but associated with a specific MACD_dS distance. It is presented on histograms for individual displacement intervals (Fig 5 –Tr3 A) where one can see a decrease of COP displacements in the firs displacement interval (0–2 mm interval at -1 second—Fig 5 Tr3, arrow I) and increase in the third interval (4–6 mm interval at -1 second—Fig 5 Tr3, arrow II). The observed differences indicate how participants tried to prepare for the upcoming move of the platform.

Taking into account previous analyses of authors and results available in other publications concerning the work of muscles [2,8,34], it can be assumed that the increase in muscle activity one second before the disturbance, resulting in an increase in the values of forces acting on the entire system, leads to an increase in the acceleration of COP movement and, consequently, an increase in the length of individual TCI_dS displacements without changes in TCI_dT. Possible correlation of the COP movement with the beginning of the platform movement—the direction of the COP movement consistent with the direction of the platform movement—could help maintain balance. The previous authors' research [6] proved however that greater COP movements after disturbance were noted in trials when the subject knew the time and type of the introduced disturbance. This may indicate that the subject was more destabilized in those trials with known parameters of the disorder compared to trials when the time and type of the disorder were not known.

It can be assumed that awareness of the approaching movement of the platform and the resulting loss of balance leads to an increase in muscle tension, which only results in stiffening of the entire body. Such stiffening can be useful when one have to counteract the impact of an incoming object but does not help in the maintaining the balance on moving basis [35,36]. These results may support everyday observations that lack of awareness of an upcoming event results in fewer injuries in events such as falls or traffic accidents. Nevertheless, the above-presented conclusions require confirmation in further research.

Conclusions

The article presents the possibilities of using the newly developed method of determining trend changes in COP movements in the study of PA phenomena as a complementary method to other COP analyses as well as EMG-based measurements. Determination of TCI parameters can show changes in the number of posture correcting movements. TCI_dS and TCI_dT as kinematic parameters can be used to explain other changes observed during such measurement, like increase in COP velocity, showing mechanism of the strategy of maintaining balance.

A significant novelty resulting from the use of the proposed method of analysis is information about the cause of changes in COP, which may result from a change in the number of posture corrections in time or from a change in the displacement between these posture corrections. Observation of changes in displacements in histograms not only allows for the detection of PA, it can also be a useful tool for measuring readiness in response to a stimulus, e.g. in sports training.

Limitations

The proposed algorithm in the current version is not sensitive to very slow postural corrections. This is the result of a very short (on the order of single seconds in this study) time window and values of parameters in stock market algorithms used to calculate TCI values. The obtained results also require a thorough medical analysis, which could indicate links to the neurological aspect related to the strategies of maintaining balance.

Supporting information

S1 File. TCIres. Results of TCI calculation.
(XLSX)

S2 File. APA. Results of APA calculation.
(XLSX)

S3 File. dSTab. Analysis of TCI_dS histograms (MAT).
(XLS)

S4 File. EPA. Results of EPA calculation.
(XLSX)

S5 File. List of additional materials with description.
(PDF)

S6 File. Mmain 4D for dS. Matlab script to 4D charts visualizations (M).
(TXT)

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RESEARCH **Open Access**

Trend change analysis of postural balance in Parkinson's disease discriminates between medication state

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Abstract

Background Maintaining static balance is relevant and common in everyday life and it depends on a correct intersegmental coordination. A change or reduction in postural capacity has been linked to increased risk of falls. People with Parkinson's disease (pwPD) experience motor symptoms affecting the maintenance of a stable posture. The aim of the study is to understand the intersegmental changes in postural sway and to apply a trend change analysis to uncover different movement strategies between pwPD and healthy adults.

Methods In total, 61 healthy participants, 40 young (YO), 21 old participants (OP), and 29 pwPD (13 during medication off, PDoff; 23 during medication on, PDon) were included. Participants stood quietly for 10 s as part of the Short Physical Performance Battery. Inertial measurement units (IMU) at the head, sternum, and lumbar region were used to extract postural parameters and a trend change analysis (TCA) was performed to compare between groups.

Objective This study aims to explore the potential application of TCA for the assessment of postural stability using IMUs, and secondly, to employ this analysis within the context of neurological diseases, specifically Parkinson's disease.

Results Comparison of sensors locations revealed significant differences between head, sternum and pelvis for almost all parameters and cohorts. When comparing PDon and PDoff, the TCA revealed differences that were not seen by any other parameter.

Conclusions While all parameters could differentiate between sensor locations, no group differences could be uncovered except for the TCA that allowed to distinguish between the PD on/off. The potential of the TCA to assess disease progression, response to treatment or even the prodromal PD phase should be explored in future studies.

Trial registration The research procedure was approved by the ethical committee of the Medical Faculty of Kiel University (D438/18). The study is registered in the German Clinical Trials Register (DRKS00022998).

Keywords Parkinson Disease, Trend Change Index, Body balance, Postural Stability, Balance, Wearable sensors, Neurology

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Introduction

Maintaining an upright posture, or static balance, is a fundamental aspect of human life that underscores the intricate interconnections of the vestibular, visual, and somatosensory systems within the central nervous system [1]. Posture is more than the mere static alignment of body segments; it represents a dynamic process characterized by continuous adjustments to maintain stability while performing various tasks. Maintaining upright posture becomes increasingly critical with aging and neurological disorders due to the gradual decline in postural control, predisposing individuals to an elevated risk of falls and associated injuries. This decline is influenced by a multitude of factors, encompassing alterations in sensory input, muscle strength, joint flexibility, and neural processing [2]. As an example pwPD present profound challenges to postural control [3] which is based on the neurodegenerative character of the disease characterized by the loss of dopaminergic neurons. The difficulties with balance are linked to the loss of dopaminergic neurons affecting the basal ganglia which are essential to control upright posture.

A particularly intriguing aspect of postural control is the necessity for specific body segments to remain stable while others adapt to accommodate external demands. For instance, the head must remain stable to preserve visual focus and spatial orientation [4], while the pelvis may need to make adjustments to accommodate changes in terrain or task requirements [5]. Unconsciously, humans stabilize their visual focus or gaze and maintain awareness of their body position [6] but also stabilize their head to ensure balance [7]. For example Wallard et al. [8] found that children with cerebral palsy exhibit greater head angle variability, suggesting a compensatory strategy and Pozzo et al. [5] observed significant head stabilization during various locomotor tasks, with the head compensating for translation and rotation. People with mild traumatic brain injury revealed increased sway of the center of mass and less head stabilization compared with healthy controls [9]. In addition Israeli-Korn et al. [10] showed that intersegmental coordination patterns differ e.g. between Parkinson's disease and cerebellar ataxia. Honegger et al. [11] investigated the coordination of the head with respect to the trunk, pelvis, and lower leg during quiet stance after vestibular loss. They argue that such simplification, as proposed by Fitzpatrick et al. [12] and Pinter et al. [13], may not fully capture the complexity of postural control in these populations. Contrary to expectations, their findings reveal synchronous movements of the head and trunk among healthy controls, suggesting that the presence of an intact vestibular system does not necessarily confer greater stability to the head in space. Instead, the pelvis emerges as a key stabilizing factor, as supported by earlier studies [13, 14] and the present investigation. These studies collectively highlight the role of aligning of body segments in postural control, particularly in individuals with motor impairments introducing another layer of complexity to our understanding of static balance. This raises the question of how the body segments sway and are controlled within the realm of quiet stance in different pathologies.

Inertial measurement units (IMUs) are small body-mounted sensors containing accelerometers, gyroscopes and magnetometers that can track 3D human movement on a very granular level e.g. to measure balance [15, 16] based on center of mass movements [17, 18]. Their reliability and validity have been extensively examined [19, 20] and provide a tool to be used in combination with a trend change analysis (TCA) [21]. TCA can detect the small number of quick corrections, an increased frequency of longer-duration corrections, and an elongation in the displacement between successive postural corrections. Adapted from techniques originally employed in stock exchange analyses, the TCA facilitates the quantification of postural corrections in both the anteroposterior (A/P) and mediolateral (M/L) directions. Moreover, it allows for the calculation of the number of adaptations, the time interval between successive posture corrections [21] providing insights about the body's responses to postural challenges [22].

The research presented herein aims to delve into the intricate relationship between maintaining an upright posture, PD, aging, and the dynamic adjustments involving intersegmental control. The objectives of this study are twofold: Firstly, to explore the potential application of TCA for the assessment of postural stability using IMUs, and secondly, to employ this analysis within the context of neurological diseases, specifically PD. We hypothesized that the TCA could differentiate between persons with PD (pwPD) and healthy adults and also distinguish, in pwPD, between dopaminergic on (PDon) and dopaminergic off phases (PDoff).

Methods

Participants

The experimental groups consisted of 61 healthy participants, 40 young (YO), 21 old (OP) and 29 pwPD. The demographic characteristics of the study participants are presented in Table 1.

All participants were either inpatients at the neurogeriatric ward of the Neurology Center at the University Hospital Schleswig-Holstein, Campus Kiel, or spouses of the patients or members of the professional team. pwPD were diagnosed according to the Movement Disorder Society clinical diagnostic criteria for Parkinson's disease [23, 24]. Thirteen pwPD participated as PDoff (UPDRS III score 24±10), 23 as PDon (UPDRS III score 30±20), and 7 as both PDon (UPDRS III score 26±10) and PDoff

Table 1 Characteristics of study participants (YO: young, OP: old, pwPD: persons with PD, w: women, m: men)

	YO	OP	pwPD
N (w/m)	40 (20/20)	21 (11/10)	29 (18/11)
Age(w/m) [year]	29.5 ± 8.5 / 27.5 ± 7.1	72.5 ± 5.9 / 70.9 ± 6.0	63.2 ± 11.7 / 68.0 ± 7.3
Weight (w/m) [kg]	79.5 ± 11.5 / 66.3 ± 8.5	83.9 ± 13.3 / 68.9 ± 12.5	88.5 ± 15.3 / 69.3 ± 14.4
Height (w/m) [m]	1.85 ± 0.08 / 1.73 ± 0.05	1.81 ± 0.08 / 1.66 ± 0.06	1.78 ± 0.07 / 1.67 ± 0.06

(UPDRS III score 27 ± 10). The sample size for this study was predetermined based on prior research and the current analysis is a secondary analysis of the previously published data set [25–27].

The study was conducted according to the guidelines of the Declaration of Helsinki and approved by the Ethics Committee of Kiel University (D438/18) and all participants provided written informed consent before participation. Participants were excluded when their fall risk was determined to be too high (>2 falls in the previous week), corrected visual acuity was below 60%, they scored ≤ 15 points in the Montreal Cognitive Assessment (MoCA) test [24, 28], had current or past chronic substance abuse (except nicotine), and were not able to perform at least one of the walking tasks [25].

Protocol

Data from the IMU sensors were recorded using a motion capture system (Noraxon USA Inc., myoMOTION 3.16, Scottsdale, AZ, USA) [25, 26]. The participants were asked to stand in an upright position with their feet together, side-by-side and fix their gaze on a point on a white wall for 10 s as part of the Short Physical Performance Battery [25].

Three IMUs were attached to the body (pelvic, sternum and head) using elastic bands with a special housing for the IMU to clip into (see Fig. 1). The research procedure was approved by the ethical committee of the Medical Faculty of Kiel University (D438/18). The study is registered in the German Clinical Trials Register (DRKS00022998).

Sensor data processing

The IMU data was processed by custom written scripts using MATLAB (MathWorks, Nantick, MA) based on methodology described by Mancini et al. [29]. The parameters provided information about the sway jerkiness (JERK) (cm^2/s^5), the sway area (SURFACE) (cm^2), path (PATH) (cm), mean velocity (MV) (cm/s), range of acceleration (RANGE) (cm/s^2) and root mean square of the acceleration (RMS) (cm/s^2).

In addition, the TCA was applied. Acceleration signals were filtered with a low-pass filter (7 Hz low-pass Butterworth filter). The method is based on a Moving Average Convergence Divergence (MACD) indicator calculation algorithm and evaluates the relationships of exponential moving averages (EMAs) for the recorded signal [21]. Calculations can be performed for any time-varying signal. In the case of the tests used, recorded acceleration signals were used, the S signal is the acceleration signal.

In the first step of calculations, for the signal S, the MACD line was determined as the difference between two EMAs (Eq. 2) with lengths of 12 and 26 samples according to Eq. 1.

$$MACD = EMA_{S,12} - EMA_{S,26} \quad \text{(Eq. 1)}$$

Where $EMA_{S,12}$ - faster exponential moving average for signal S,

$EMA_{S,26}$ - slower exponential moving average for signal S

$$EMA = \frac{p_0 + (1-\alpha)\,p_1 + (1-\alpha)^2 p_2 + \cdots + (1-\alpha)^N p_N}{1 + (1-\alpha) + (1-\alpha)^2 + \cdots + (1-\alpha)^N} \quad \text{(Eq. 2)}$$

Where, p_0 – ultimate value, p_1 – penultimate value, p_N – value preceding N periods, N = number of periods, α = a smoothing coefficient equal to $2/(N+1)$.

In the next step, the signal line is calculated as an EMA with a length of 9 samples from the MACD line signal in accordance with Eq. 3.

$$Signal\ line = EMA_{MACDline,9} \quad \text{(Eq. 3)}$$

The intersection of the MACD line and the Signal line determines the trend change points in the S signal. The number of intersections determines the TCI (trend changes index).

In the next step, the time intervals between successive points of trend changes in the S signal were calculated. In this way, the MACD_dT array was determined, the average value of which is the value of the TCI_dT. As a consequence, the displacement between subsequent trend change points were calculated and the results constitute the MACD_dS array. The average value of the array is the value of the TCI_dS (Fig. 2). Finally, the corresponding elements of the MACD_dS array were divided by MACD_dT to obtain the MACD_dV array. The average value of the array is the value of the TCI_dV. In this study, the displacement of the signal is the difference in the acceleration values between successive points of trend change on the acceleration signal.

To summarize, TCI determines the number of trend changes in the assumed research period, TCI_dT defines the average time between detected trend changes, and TCI_dS determines the average value of the acceleration change between subsequent trend changes. Indices were

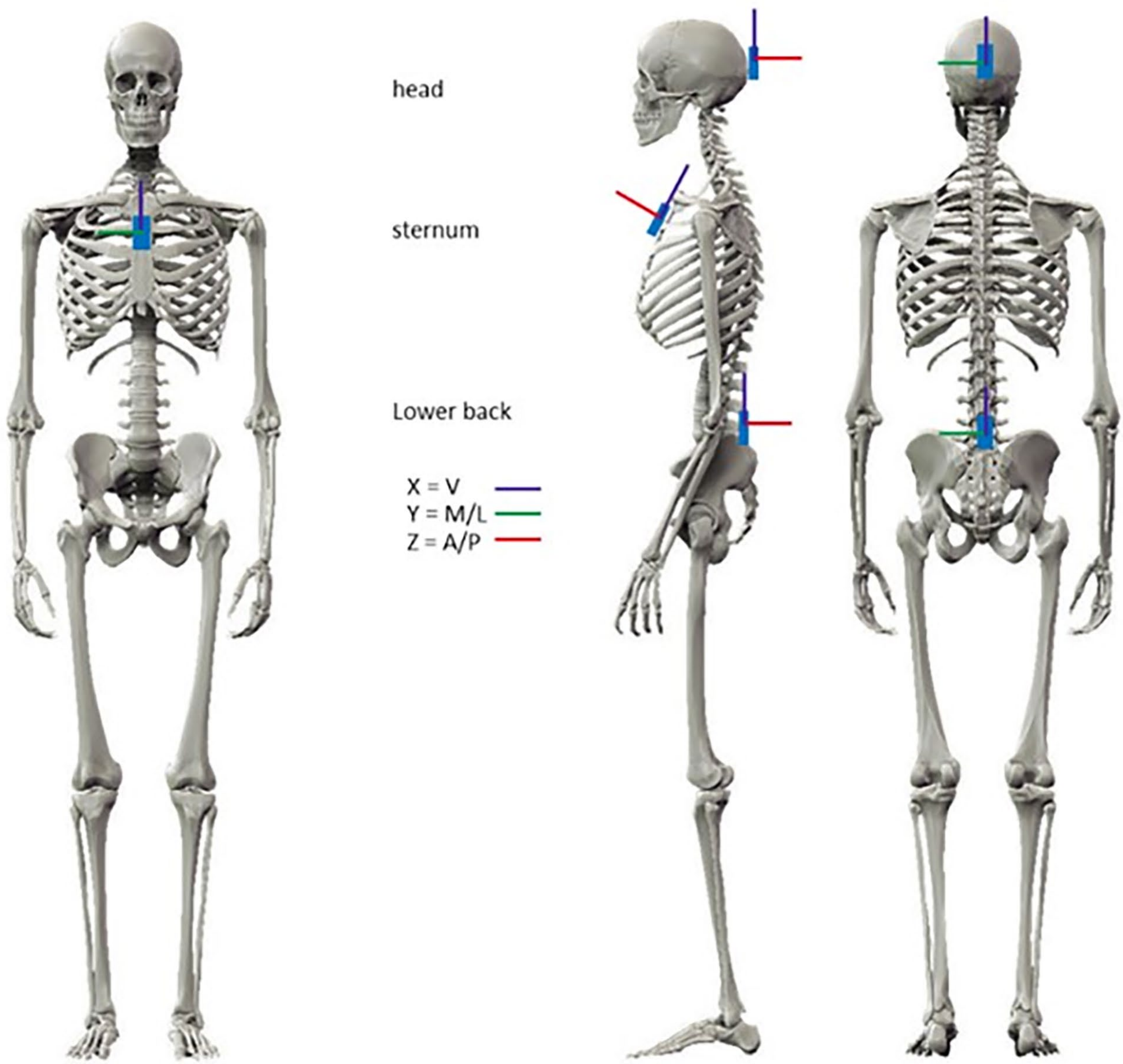

Fig. 1 Placement of the inertial measurement units on the head, sternum and pelvis

determined for each of the three directions of measurement, and then the resultant values were determined i.e. for TCI as the sum of the number of trend changes detected in each direction of the measured accelerations (in the X, Y and Z axes), and for TCI_dT, TCI_dS, TCI_dV as the square root of the sum of squares of the values calculated in each direction.

Statistical analysis

The analyses were performed using Matlab R2022a and JASP (Version 0.16.1 JASP Team (2022)) for all statistical analyses.

The analysis aimed to investigate differences between sensor positions and cohorts within the dataset. Shapiro-Wilk tests revealed significant deviations from normality ($p<0.05$) across multiple groups and sensor positions, thus prompting the utilization of non-parametric tests. Subsequently, a Kruskal-Wallis H Test were employed to evaluate variations between cohorts and sensor positions. In case of statistically significant differences ($p<0.05$) post-hoc analyses, utilizing Dunn's test with Bonferroni correction, were conducted to ascertain specific group disparities.

Results

When comparing the individual parameters for each sensor and each cohort (Table 2), no differences could be found between the cohorts but significant differences

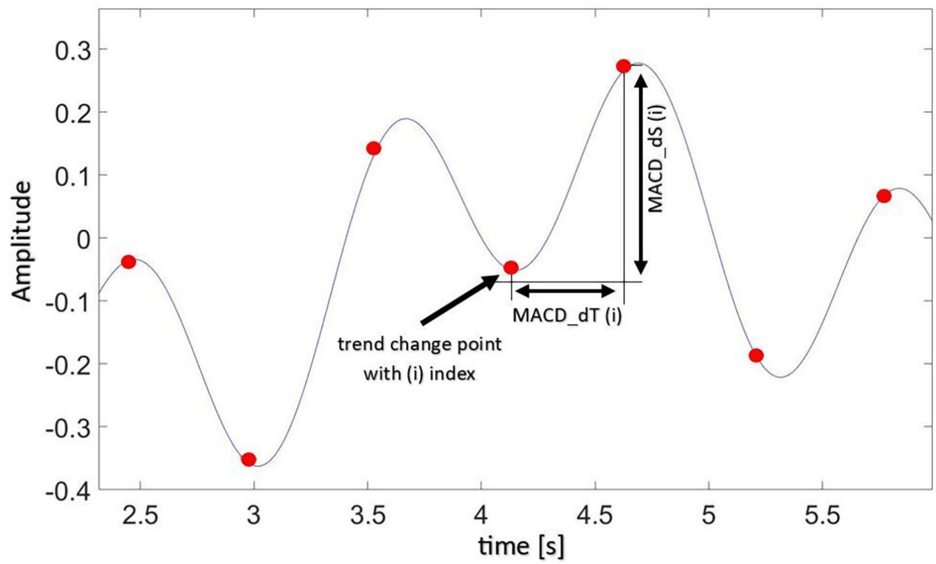

Fig. 2 Graphical explanation of the Trend Change Index (TCI), the delta time between successive TCIs (MACD_dT) as well as the delta space between successive TCIs (MACD_dS) in an acceleration signal from a sensor on the pelvis with an observation phase of about 3 s. Seven trend changes (indicated by the seven red dots) are shown. All determined MACD_dTs were used to calculate TCI_dT and all MACD_dSs to calculate TCI_dS according to the procedure described in the text

were uncovered between the sensor positions (Additional file 1).

The sensor position differed for all cohorts and all parameters except TCI and TCI_dT for PDon (Table 3).

When comparing the PDon and PDoff cohort (Table 4) only TCI & TCI_dT differed between the PDon and PDoff cohort. Significant differences were found between the three sensor locations (Table 5).

Discussion

This study investigated postural stability of healthy young, old controls and persons with PD in a static balance task using three different sensor locations. The aim of the study was to analyze the upright posture and intersegmental adjustments, to evaluate whether the parameters could uncover distinct postural sway behavior between the different cohorts. Our results confirmed that both, the postural parameters and TCA, could uncover sway differences between the segments but only the TCA could differentiate between PDon and PDoff.

The results of the current study show no group differences between the healthy adults and pwPD, confirming results from a previous study investigating static sway with increasing task difficulty [27]. This is of interest as PD is known for its altered postural reflexes with a disruption of the precisely coordinated execution of agonist and antagonist muscles (associated with bradykinesia and rigidity), which results in difficulty to maintain static postural stability [30–32] due to a reduced margin of stability [33].

While pwPD have shown larger values for sway acceleration, jerk and sway velocity during postural balance compared to age-matched healthy controls [29, 34] they also show an increased jerkiness during the performance of cognitive task [35], suggesting an interaction of cognitive functions, including multisensory integration, with static balance mechanisms. Our results highlight larger motions from the head compared to the sternum and the pelvis. The results convey with previous findings [14] basing their findings upon the biomechanical principal of a double-inverted pendulum. The double-inverted pendulum allows to be controlled by the ankles, the hip or both, while assuming a rigid head-on-trunk coupling. Almost all parameters were able to distinguish between sensor position indicating the complex relationship between the dynamic intersegmental adjustments and upright posture. The results suggest that for a relative simple and short balance tasks pwPD can perform control-like, which could be related to the location of the pathology within the central nervous system and its extensive compensation possibilities [36] and by using alternative pathways or even networks [37].

There is some evidence that dopaminergic medication can improve static sway [38, 39]. However, there are not many IMU-based studies available that can show these differences. One reason may be that the parameters currently assessed for this performance are not covering disease-relevant changes. Here we introduced TCA in the analysis of static sway in PDon and PDoff, and could in fact detect significant differences only with this approach (but not with the conventional parameters). We found

Table 2 Values of individual parameters (M - median, QR/2 - the half of coefficient quartile of variation)

		Head				Pelvis				Sternum			
		YO	**OP**	**PD on**	**PD off**	**YO**	**OP**	**PD on**	**PD off**	**YO**	**OP**	**PD on**	**PD off**
JERK [cm2/s5]	**M**	30.78	37.54	28.52	28.06	7.64	10.48	8.28	8.08	9.22	12.12	10.26	8.59
	QR/2 [%]	190	123	56	60	58	54	67	59	109	58	55	50
MV	**M**	24.17	29.66	26.50	26.97	12.15	13.32	14.39	15.57	14.25	16.78	16.80	15.98
[cm/s]	**QR/2 [%]**	38	29	22	24	31	34	29	31	31	21	17	24
PATH	**M**	241.69	296.59	265.04	269.73	121.47	133.16	143.92	155.67	142.52	167.84	168.02	159.75
[cm]	**QR/2 [%]**	38	29	22	24	31	34	29	31	31	21	17	24
RMS [cm/s2]	**M**	2.72	3.18	2.11	3.13	0.91	1.21	1.14	1.17	1.15	1.47	1.43	1.34
	QR/2 [%]	67	72	83	36	25	34	38	14	44	41	38	29
SURFACE [cm2/s4]	**M**	35.96	81.86	35.94	71.10	7.09	10.40	10.90	11.22	10.49	18.07	17.19	14.17
	QR/2 [%]	229	145	209	74	49	77	48	29	89	84	70	61
RANGE	**M**	10.73	12.02	7.83	9.38	2.97	4.29	4.19	3.38	3.43	5.61	4.37	4.46
[cm/s2]	**QR/2 [%]**	78	89	64	73	41	52	44	15	72	53	43	24
TCI	**M**	258	272	269	302	295	295	284	317	296	301	287	310
[no]	**QR/2 [%]**	9	5	4	4	4	6	7	5	3	5	7	2
TCI_dT	**M**	0.21	0.20	0.19	0.17	0.18	0.18	0.18	0.16	0.18	0.17	0.18	0.17
[s]	**QR/2 [%]**	8	4	5	4	4	6	9	5	4	5	6	2
TCI_dS [cm]	**M**	2.08	2.29	1.89	1.76	0.88	0.74	1.02	0.97	1.01	0.96	1.02	1.02
	QR/2 [%]	59	40	35	23	36	46	44	27	35	36	35	18
TCI_dV	**M**	15.63	19.49	15.90	17.22	8.26	6.90	9.59	9.20	9.32	8.45	11.18	10.04
[cm/s]	**QR/2 [%]**	48	32	30	25	34	56	39	26	35	44	25	19

Table 3 Sensor parameters to differentiate between groups and sensor positions in controls and PDon. The H-statistics of the Kruskall-Wallis test as well as the degree of freedom and significance levels are reported within the tables

Parameters	Group level	Sensor position	YO post hoc $p<0.05$	OP post hoc $p<0.05$	PDon post hoc $p<0.05$
JERK	n.s.	H(2) = 60.29, $p<0.001$	head vs. sternum and pelvis	head vs. sternum and pelvis	head vs. sternum and pelvis
MV	n.s.	H(2) = 70.87, $p<0.001$	head vs. sternum and pelvis	head vs. sternum and pelvis	head vs. sternum and pelvis
PATH	n.s.	H(2) = 70.87, $p<0.001$	head vs. sternum and pelvis	head vs. sternum and pelvis	head vs. sternum and pelvis
RMS	n.s.	H(2) = 73.18, $p<0.001$	head vs. sternum and pelvis	head vs. sternum and pelvis	head vs. sternum and pelvis
SURFACE	n.s.	H(2) = 69.59, $p<0.001$	head vs. sternum and pelvis	head vs. pelvis	head vs. sternum and pelvis
RANGE	n.s.	H(2) = 54.82, $p<0.001$	head vs. sternum and pelvis	head vs. pelvis	head vs. sternum and pelvis
TCI	n.s.	H(2) = 44.27, $p<0.001$	head vs. sternum and pelvis	head vs. sternum and pelvis	
TCI_dT	n.s.	H(2) = 57.37, $p<0.001$	head vs. sternum and pelvis	head vs. sternum and pelvis	
TCI_dS	n.s.	H(2) = 79,63, $p<0.001$	head vs. sternum and pelvis	head vs. sternum and pelvis	head vs. sternum and pelvis
TCI_dV	n.s.	H(2) = 58.94, $p<0.001$	head vs. sternum and pelvis	head vs. sternum and pelvis	head vs. sternum and pelvis

Table 4 Values of parameter for 7 pwPD tested "on" and "off" (M - median, QR/2 - the half of coefficient quartile of variation)

		Head		Pelvis		Sternum	
		PD off	**PD on**	**PD off**	**PD on**	**PD off**	**PD on**
JERK [cm2/s5]	**M**	34.71	44.54	13.08	8.07	8.87	10.26
	QR/2 [%]	67	27	79	6	267	39
MV [cm/s]	**M**	32.82	29.79	18.66	14.39	16.57	16.80
	QR/2 [%]	17	15	28	6	59	10
PATH [cm]	**M**	328.20	297.90	186.62	143.92	165.71	168.02
	QR/2 [%]	17	15	28	6	59	10
RMS [cm/s2]	**M**	3.27	3.67	1.23	1.53	1.84	1.69
	QR/2 [%]	21	51	11	21	28	16
SURFACE [cm2/s4]	**M**	77.93	113.99	11.33	12.86	27.54	24.67
	QR/2 [%]	55	81	31	21	44	35
RANGE [cm/s2]	**M**	13.87	10.17	3.35	4.27	5.20	5.75
	QR/2 [%]	40	83	15	33	21	14
TCI [no]	**M**	308	275	316	279	310	277
	QR/2 [%]	3	3	4	6	2	3
TCI_dT [s]	**M**	0.17	0.19	0.16	0.19	0.17	0.19
	QR/2 [%]	4	3	5	5	2	3
TCI_dS [cm]	**M**	1.85	2.71	1.11	1.05	1.10	1.27
	QR/2 [%]	21	26	26	18	26	23
TCI_dV [cm/s]	**M**	20.48	17.68	11.55	9.77	10.68	11.18
	QR/2 [%]	15	27	32	8	27	19

a higher number of TCIs and smaller TCI_dT values in PDoff compared to PDon. This is coherent with previous results obtained for COP measurements showing an increase in TCIs and reduction of TCI_dT in pwPD compared to healthy individuals [40]. In our view, this perspective also aligns with a pathomechanistic standpoint. Previous research, as indicated by Bizid et al. [41], suggests that low frequencies are predominantly associated with visuo-vestibular regulation, while high frequencies are associated with proprioceptive regulation. Additionally, it is well-established that visual perception and integration are strongly dopamine-dependent [42]. Therefore, we hypothesize that the results observed through TCA most likely reflect visual deficits resulting from a dopaminergic deficit. This is particularly evident,

Table 5 Sensor parameters to differentiate between groups and sensor positions in PDon and PDoff. The H-statistics of the Kruskall-Wallis test as well as the degree of freedom and significance levels are reported within the tables

Parameters	Group level	Sensor position
JERK	n.s.	$H(2)=12.63, p=0.002$
MV	n.s.	$H(2)=11.11, p=0.004$
PATH	n.s.	$H(2)=11.11, p=0.004$
RMS	n.s.	$H(2)=13.09, p=0.001$
SURFACE	n.s.	$H(2)=17.12, p<0.001$
RANGE	n.s.	$H(2)=11.59, p=0.003$
TCI	$H(2)=13.40, p<0.001$	n.s.
TCI_dT	$H(2)=13.21, p<0.001$	n.s.
TCI_dS	n.s.	$H(2)=9.13, p=0.01$
TCI_dV	n.s.	$H(2)=7.59, p=0.022$

given that lower leg proprioceptive performance does not appear to be influenced by dopaminergic treatment [43].

Limitations

It would be worthwhile to mention limitations of the current study. First, the number of pwPD measured in both medication states was relatively low, potentially limiting the generalizability of findings and the ability to capture the full spectrum of balance-related issues in PD. Another constraint lies in the brief 10-second measurement duration, which may not provide a comprehensive representation of individuals' balance control capabilities, particularly in dynamic real-world scenarios. Additionally, the use of a side-by-side stance as a measure may cause limitations as it may not be challenging enough to detect subtle differences between cohorts or uncover changes in postural control based on intersegmental coordination. These limitations emphasize the need for cautious interpretation of results and highlight areas for future research to address these constraints and provide a more nuanced understanding of balance control in Parkinson's disease and other relevant populations. Nevertheless, considering these limitations, it is all the more remarkable given that the TCA parameters were effective in distinguishing between PD on and PD off.

Clinical implication

This study investigated static sway in healthy individuals and pwPD using three sensor locations. Results show that postural parameters effectively distinguish between segments. However, and even more relevant, the introduction of TCA proves instrumental in detecting significant differences between PDon and off medication, showcasing its potential in assessing disease-relevant changes not captured by conventional parameters.

Supplementary Information

The online version contains supplementary material available at https://doi.org/10.1186/s12984-024-01411-z.

Supplementary Material 1

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Author contributions
RR and MAH and WM and CH and EW and MCh and PW and JJ: made the conception RR and MAH and EW and CH: data acquisitionPW and JJ and MCh and CH: analysisPW and WM and JJ and CH: interpretation of dataPW and JJ and KC: creation of new software used in the workWM and JJ and PW and CH: have drafted the work PW and JJ and RR and MAH and WM and CH and EW and MCh and KC: substantively revised the manuscrypt.

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Data availability
No datasets were generated or analysed during the current study.

Declarations

Ethics approval and consent to participate
The research procedure was approved by the ethical committee of the Medical Faculty of Kiel University (D438/18). The study is registered in the German Clinical Trials Register (DRKS00022998).

Consent for publication
All authors express their full consent to publication of the material.

Competing interests
The authors declare no competing interests.

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Publisher's Note

Oświadczenia o indywidualnym wkładzie procentowym i merytorycznym współautorów w powstanie pracy zbiorowej

Gliwice, 23.09.2024 r.

Oświadczenie o indywidualnym wkładzie procentowym i merytorycznym współautorów w powstanie pracy zbiorowej

Publikacja:

[A1] Wodarski Piotr, Jurkojć Jacek, Chmura Marta, Gruszka Grzegorz, Gzik Marek. Analysis of center of pressure displacements and head movements triggered by a visual stimulus created using the virtual reality technology. *Acta Bioeng Biomech*. 2022;24(1):19-28. DOI: 10.37190/ABB-01900-2021-01. Punktacja MNiSW = 100. IF$^{(2022)}$ = 1,000.

Wykaz autorów wraz z udziałem procentowym:

Lp.	Imię i nazwisko	Udział procentowy
1	Piotr Wodarski *	50%
2	Jacek Jurkojć	25%
3	Chmura Marta	15%
4	Grzegorz Gruszka	5%
5	Marek Gzik	5%

* Autor korespondencyjny

Wkład merytoryczny doktorantki:

- Przygotowanie stanowiska badawczego
- Przeprowadzenie eksperymentu
- Udział w przygotowaniu manuskryptu

Podpisał prof. Piotr Wodarski (podpis odręczny)

Podpis pierwszego autora / autora korespondencyjnego

Gliwice, 23.09.2024 r.

Oświadczenie o indywidualnym wkładzie procentowym i merytorycznym współautorów w powstanie pracy zbiorowej

Publikacja:

[A2] Wodarski Piotr, Chmura Marta, Szlęzak Michał, Gruszka Grzegorz, Romanek Justyna, Jurkojć Jacek. The effect of selected lower limb muscle activities on a level of imbalance in reaction on anterior-posterior ground perturbation. *Acta Bioeng Biomech*. 2022;24(3):135-146. DOI: 10.37190/ABB-02112-2022-02. Punktacja MNiSW = 100. IF$^{(2022)}$ = 1,000.

Wykaz autorów wraz z udziałem procentowym:

Lp.	Imię i nazwisko	Udział procentowy
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2	Marta Chmura	20%
3	Michał Szlęzak	10%
4	Grzegorz Gruszka	5%
5	Justyna Romanek	5%
6	Jacek Jurkojć	20%

* Autor korespondencyjny

Wkład merytoryczny doktorantki:

- Przygotowanie stanowiska badawczego
- Przeprowadzenie eksperymentu
- Udział w organizacji i analizie wyników
- Udział w przygotowaniu manuskryptu

Podpisał prof. Piotr Wodarski (podpis odręczny)

Podpis pierwszego autora / autora korespondencyjnego

Gliwice, 23.09.2024 r.

Oświadczenie o indywidualnym wkładzie procentowym i merytorycznym współautorów w powstanie pracy zbiorowej

Publikacja:

[A3] Wodarski Piotr, Chmura Marta, Gruszka Grzegorz, Romanek Justyna, Jurkojć Jacek. The stock market indexes in research on human balance. *Acta Bioeng Biomech.* 2022;24(2):163-176. DOI: 10.37190/ABB-02062-2022-03. Punktacja MNiSW = 100. IF$^{(2022)}$ = 1,000.

Wykaz autorów wraz z udziałem procentowym:

Lp.	Imię i nazwisko	Udział procentowy
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3	Grzegorz Gruszka	5%
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* Autor korespondencyjny

Wkład merytoryczny doktorantki:

- Przygotowanie stanowiska badawczego
- Przeprowadzenie eksperymentu
- Organizacja i analiza wyników
- Udział w przygotowaniu manuskryptu

Podpisał prof. Piotr Wodarski (podpis odręczny)

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Gliwice, 23.09.2024 r.

Oświadczenie o indywidualnym wkładzie procentowym i merytorycznym współautorów w powstanie pracy zbiorowej

Publikacja:

[A4] Wodarski Piotr, Chmura Marta, Jurkojć Jacek. Impact of Visual Disturbances on the Trend Changes of COP Displacement Courses Using Stock Exchange Indices. *Applied Sciences*. 2024; 14(11):4953. DOI: 10.3390/app14114953. Punktacja MNiSW = 100. IF$^{(2023)}$ = 2,500.

Wykaz autorów wraz z udziałem procentowym:

Lp.	Imię i nazwisko	Udział procentowy
1	Piotr Wodarski	50%
2	Marta Chmura	40%
3	Jacek Jurkojć *	10%

* Autor korespondencyjny

Wkład merytoryczny doktorantki:

- Przygotowanie stanowiska badawczego
- Przeprowadzenie eksperymentu
- Udział w organizacji i analizie wyników
- Analiza statystyczna wyników
- Udział w opracowywaniu dyskusji i wniosków
- Udział w przygotowaniu manuskryptu

Podpisał prof. Piotr Wodarski (podpis odręczny)

Podpis pierwszego autora / autora korespondencyjnego

Gliwice, 23.09.2024 r.

Oświadczenie o indywidualnym wkładzie procentowym i merytorycznym współautorów w powstanie pracy zbiorowej

Publikacja:

[A5] Wodarski Piotr, Chmura Marta, Szlęzak Michał, Bajor Grzegorz, Gzik Marek, Jurkojć Jacek. Trend change analysis in the assessment of body balance during posture adjustment in reaction to anterior-posterior ground perturbation. *PLoS One*. 2024;19(4):e0301227. DOI: 10.1371/journal.pone.0301227. Punktacja MNiSW = 100. $IF^{(2023)}$ = 2,900.

Wykaz autorów wraz z udziałem procentowym:

Lp.	Imię i nazwisko	Udział procentowy
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2	Marta Chmura *	15%
3	Michał Szlęzak	5%
4	Grzegorz Bajor	5%
5	Marek Gzik	5%
6	Jacek Jurkojć	20%

* Autor korespondencyjny

Wkład merytoryczny doktorantki:

- Przygotowanie stanowiska badawczego
- Przeprowadzenie eksperymentu
- Organizacja wyników
- Udział w przygotowaniu manuskryptu

Podpisał prof. Piotr Wodarski (podpis odręczny)

Podpis pierwszego autora / autora korespondencyjnego

Gliwice, 23.09.2024 r.

Oświadczenie o indywidualnym wkładzie procentowym i merytorycznym współautorów w powstanie pracy zbiorowej

Publikacja:

[A6] Wodarski Piotr, Jurkojć Jacek, Chmura Marta, Elke Warmerdam, Robbin Romijnders, Markus A Hobert, Walter Maetzler, Krzysztof Cygoń, Clint Hansen. Trend change analysis of postural balance in Parkinson's disease discriminates between medication state. *J Neuroeng Rehabil.* 2024;21(1):112. Published 2024 Jun 28. DOI: 10.1186/s12984-024-01411-z. Punktacja MNiSW = 140. $IF^{(2023)}$ = 5,200.

Wykaz autorów wraz z udziałem procentowym:

Lp.	Imię i nazwisko	Udział procentowy
1	Piotr Wodarski	26%
2	Jacek Jurkojć	15%
3	Marta Chmura	15%
4	Elke Warmerdam	4%
5	Robbin Romijnders	4%
6	Markus A. Hobert	4%
7	Walter Maetzler	15%
8	Krzysztof Cygoń	2%
9	Clint Hansen	15%

* Autor korespondencyjny

Wkład merytoryczny doktorantki:

- Analiza statystyczna wyników
- Udział w opracowywaniu dyskusji i wniosków
- Udział w przygotowaniu manuskryptu

Podpisał prof. Piotr Wodarski (podpis odręczny)

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Życiorys Autorki

Mgr inż. Marta Chmura w 2015 roku rozpoczęła studia inżynierskie na Wydziale Inżynierii Biomedycznej Politechniki Śląskiej na kierunku inżynieria biomedyczna o specjalności inżynieria wyrobów medycznych. W 2019 roku obroniła tytuł inżyniera. Następnie kontynuowała studia magisterskie na Wydziale Inżynierii Biomedycznej Politechniki Śląskiej, na kierunku inżynieria biomedyczna o specjalności biomechatronika i sprzęt medyczny. W 2020 roku obroniła pracę magisterską pt. „Analiza wpływu czasu przebywania w środowisku wirtualnej rzeczywistości na wielkości opisujące zdolność utrzymywania równowagi i zmienność chodu", uzyskując wynik bardzo dobry.

Od 1 października 2020 roku rozpoczęła dalsze kształcenie w ramach Wspólnej Szkoły Doktorskiej Politechniki Śląskiej dokonując wyboru tematu rozprawy doktorskiej zaproponowanego przez promotora, dr. hab. inż. Jacka Jurkojcia, prof. PŚ pt. „Ocena zmian wybranych mechanizmów kontroli posturalnej w odpowiedzi na rzeczywiste i wirtualne bodźce prowadzące do wytrącenia z równowagi".

Doktorantka jest współautorką 8 artykułów naukowych opublikowanych w czasopismach, w tym 7 z listy JCR. Jest również współautorką 15 rozdziałów z monografii. Wzięła udział w 14 konferencjach naukowych, krajowych oraz międzynarodowych. Pozyskała grant projakościowy na rozpoczęcie działalności naukowej w nowej tematyce badawczej, w ramach programu Inicjatywa Doskonałości – Uczelnia Badawcza oraz subwencję na utrzymanie i rozwój potencjału badawczego Młodych Naukowców. W trakcie trwania studiów doktoranckich odbyła 7-miesięczny staż naukowy w zakładzie fizjoterapii Fizjosport w Gliwicach oraz miesięczny staż naukowy, zagraniczny, na Uniwersytecie w Kilonii w Niemczech. Poszerzając zakres swoich kompetencji, wzięła udział w letniej szkole "Motor Control Summer School" we wrześniu 2024 roku oraz w szkoleniu prosektoryjno-anatomicznym w czerwcu 2023 roku. Otrzymała wyróżnienie w konkursie o Nagrodę Polskiego Towarzystwa Biomechaniki dla młodych pracowników nauki im. prof. A. Moreckiego i prof. K. Fidelusa w edycji 2023.

Doktorantka podczas studiów czynnie angażowała się w działalność promocyjną uczelni będąc od października 2020 roku czynnym członkiem Zespołu ds. Promocji i Popularyzacji Nauki na Wydziale Inżynierii Biomedycznej. Była także członkiem komitetów organizacyjnych konferencji naukowych: biorąc udział w organizowaniu konferencji naukowych: Advances in Applied Biomechanics & „Majówka Młodych Biomechaników" im. prof. Dagmary Tejszerskiej (2021 r.), "Medical and Sport Technologies" & „Young Biomechanists Conference" named of prof. Dagmara Tejszerska (2022-2024 r.) i HealthTech Innovations Conference (2022-2024 r.). W 2021 roku została powołana na stanowisko prezesa Rady Młodych Innowatorów, działającej przy Towarzystwie Innowacyjnych Technologii dla Zdrowia. Brała także udział w akcjach organizowanych przez Centrum Popularyzacji Nauki Politechniki Śląskiej, takich jak Noc Naukowców czy Dni Otwarte Politechniki Śląskiej.