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The validity and usability of an eight marker model for avatar-based biofeedback gait training

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ABSTRACT

Background: Virtual reality presents a platform for therapeutic gaming, and incorporation of immersive biofeedback on gait may enhance outcomes in rehabilitation. Time is limited in therapeutic practice, therefore any potential gait training tool requires a short set up time, while maintaining clinical relevance and accuracy. The aim of this study was to develop, validate, and establish the usability of an avatar-based application for biofeedback-enhanced gait training with minimal set up time.

Methods: A simplified, eight marker model was developed using eight passive markers placed on anatomical landmarks. This allowed for visualisation of avatar-based biofeedback on pelvis kinematics, hip and knee sagittal angles in real-time. Retrospective gait analysis data from typically developing children (n = 41) and children with cerebral palsy (n = 25), were used to validate eight marker model. Gait outcomes were compared to the Human Body Model using statistical parametric mapping. Usability for use in clinical practice was tested in five clinical rehabilitation centers with the system usability score.

Findings: Gait outcomes of Human Body Model and eight marker model were comparable, with small differences in gait parameters. The discrepancies between models were < 5°, except for knee extension where eight marker model showed significantly less knee extension, especially towards full extension. The application was considered of 'high marginal acceptability' (system usability score, mean 68 (SD 13)).

Interpretation: Gait biofeedback can be achieved, to acceptable accuracy for within-session gait training, using an eight marker model. The application may be considered usable and implemented for use in patient populations undergoing gait training.

1. Introduction

Gait impairments are one of the most commonly reported symptoms in patients with neurologic disorders (Stolze et al., 2005). Limitations in an individual's ability to ambulate without assistance present barriers to independence and performance of activities of daily living. As such, rehabilitation of gait is a primary therapeutic goal for many patients (Dobkin, 2005). Repetitive task-specific training, such as gait training can be an effective tool for improving gait related functional outcomes, across patient populations such as cerebral palsy (CP) (Booth et al., 2018; Chrysagis et al., 2012; Hamed et al., 2011), stroke (Mehrholz et al., 2014), spinal cord injuries (Schwartz and Meiner, 2015) and Parkinson's disease (Herman et al., 2007). Gait training encompasses a range of interventions and may include the use of treadmills, body-weight support and robotic assistive devices (Booth et al., 2018; Damiano et al., 2009; Lefmann et al., 2017).

Across all gait training interventions there is the possibility to incorporate the use of virtual reality (VR). There is suggestive evidence that the addition of VR in therapy may enhance intervention outcomes (Booth et al., 2018; Yang et al., 2008). With advances in technology the potential applications of VR are expanding. VR allows the patient to experience stimulating challenges, transporting them outside of the usual therapy environment. It can help to maintain motivation and engagement in prolonged therapy (Sveistrup, 2004). When applied in the context of motor learning, VR may help in facilitating task specificity, variation and motivation (Spiess et al., 2018). A potentially valuable feature of VR is the interaction the user can have with the environment, integrating aspects of purposeful gaming and feedback on performance.

Feedback is important in performance of motor tasks and

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stimulating motor learning (Sigrist et al., 2013; Sveistrup, 2004). In populations with impaired motor control, such as CP and stroke, the intrinsic sensory feedback may be diminished or maladapted. Providing additional, augmented feedback on the movement of the body during walking may act to increase the sensory information available to the individual, this is termed 'biofeedback'. Biofeedback has a diverse range of applications in rehabilitation (Giggins et al., 2013). While verbal biofeedback has been used by therapists for many years, it can be difficult to verbally convey accurate and meaningful biofeedback on specific movements such as knee extension. Integrating VR and precise visual biofeedback may present a novel therapeutic paradigm for gait training rehabilitation. It was shown that children with CP had the ability to adapt and improve a range of gait parameters (Booth et al., 2019; van Gelder et al., 2017). The method of presentation of biofeedback is important in its success as a treatment (MacIntosh et al., 2018; Richards et al., 2018).While there is currently no defined optimum, the use of a simplified avatar, representing a patient's movements in real-time has shown promise in stroke (Loudon et al., 2012; Thikey et al., 2012) and CP rehabilitation (Booth et al., 2019).

In clinical therapy, time is often a limiting factor. In standard gait analysis, subject preparation with marker placement and calibration is approximately 30 min (Leardini et al., 2007). This is not feasible for a training programme and thus acts as a barrier to the uptake of biofeedback enhanced gait re-training. Therefore, the aim of this investigation was to develop, validate and evaluate the clinical usability of a VR application to provide immersive biofeedback on gait kinematics and spatiotemporal parameters with minimal preparation time. It was expected that with eight passive markers, kinematics of the pelvis, sagittal hip and knee angles, and spatiotemporal parameters could be calculated and visualised in real-time to an acceptable level of accuracy in both healthy and pathological gait. Clinical usability was established by piloting the application at specialist rehabilitation centers.

2. Methods

2.1. Study design

The study involved a three-stage process. First, an application and eight marker model (8MM) were developed using D-Flow software (v3.28, Motek Medical B.V., Netherlands), integrating an instrumented treadmill, motion capture, VR and real-time processing of data. The application allowed for real-time calculation of basic gait outcomes and visualisation of biofeedback with an avatar (Fig. 1). Secondly, retrospective gait analysis data was used to validate the kinematic outcomes of the 8MM. Finally, the application was disseminated to five specialist rehabilitation centers to trial the application and evaluate usability in clinical practice.

Table 1
Subject demographics.

	CP (n = 25)		TD (n = 41)	
	Mean (SD)/n	Range	Mean (SD)/n	Range
Age (years)	10.4 (2.9)	6–16	10.0 (3.0)	5–15
Height (m)	1.50 (0.20)	1.27–1.9	1.47 (0.19)	1.27–1.9
Body mass (kg)	40.8 (17.3)	26.1–89	40.1 (13.2)	19.1–67.1
Walking speed (m/s)	0.65 (0.18)	0.35–1.1	1.05 (0.20)	0.6–1.37
Sex	M: 16, F: 9	–	M: 24, F: 17	–
GMFCS	I: 12, II: 13	–	–	–
Localisation	Unilateral: 6, Bilateral: 17	–	–	–

2.2. Description of participants

Retrospective data from gait analysis in 41 typically developing (TD) children and 25 children with CP was used to provide input data for the validation of 8MM. Demographics are presented in Table 1. TD children were aged between 5 and 16 and had no impairments or conditions that could interfere with walking ability. Exclusion criteria were any injuries, physical conditions or cognitive disabilities that might affect walking ability. Children with CP were included under the following criteria; diagnosis of spastic paresis (both unilateral and bilateral), walking without aids (gross motor classification system (GMFCS) level I-II) and aged between 5 and 16 years old. Children were recruited from the VU University Medical Center, Amsterdam and Revant Rehabilitation Center, Breda. Written informed consent was obtained from all parents, and children 12 years and older.

2.3. Data collection

All participants walked on a dual-belt instrumented treadmill with immersive VR environment (GRAIL, Motek Medical B.V., Amsterdam, Netherlands. Fig. 1). A 10-camera 3D motion capture (Vicon Motion Systems Ltd, Oxford, UK) system was used with 26 retroreflective markers placed on anatomical landmarks following HBM (van den Bogert et al., 2013). A non-weight bearing safety harness was worn to prevent injury in case of an accidental trip. Following a period of at

Fig. 1. Experimental set-up with treadmill and virtual reality screen (GRAIL, Motek Medical B.V., Amsterdam, Netherlands). 3D motion capture (Vicon, Oxford, UK) is integrated with virtual reality using D-Flow software (Motek Medical B.V., Amsterdam, Netherlands) to provide real-time gait analysis and visualisation. With placement of 8 markers, lower limb and trunk movements are visualised using an avatar. Intuitive biofeedback on a range of gait parameters may be attached to the avatar depending on the goal of the patient. This may include step length/width (as pictured), cadence, pelvis tilt, pelvis obliquity, hip extension and/or knee extension at initial contact. Targets for gait modification are set relative to patient baseline performance and may be adjusted by the therapist to maintain difficulty. Auditory rewards and score progression provide knowledge of results and motivation.

Marker	Placement guide
JN	On the jugular notch of the sternum.
R/L ASIS	On the prominence of the anterior superior iliac spine.
SACR	The midpoint of the posterior superior iliac spines.
R/L KNEE	Lateral epicondyle of the knee, on line of joint axis.
R/L ANK	Lateral malleolus of the ankle, at the center of prominence.

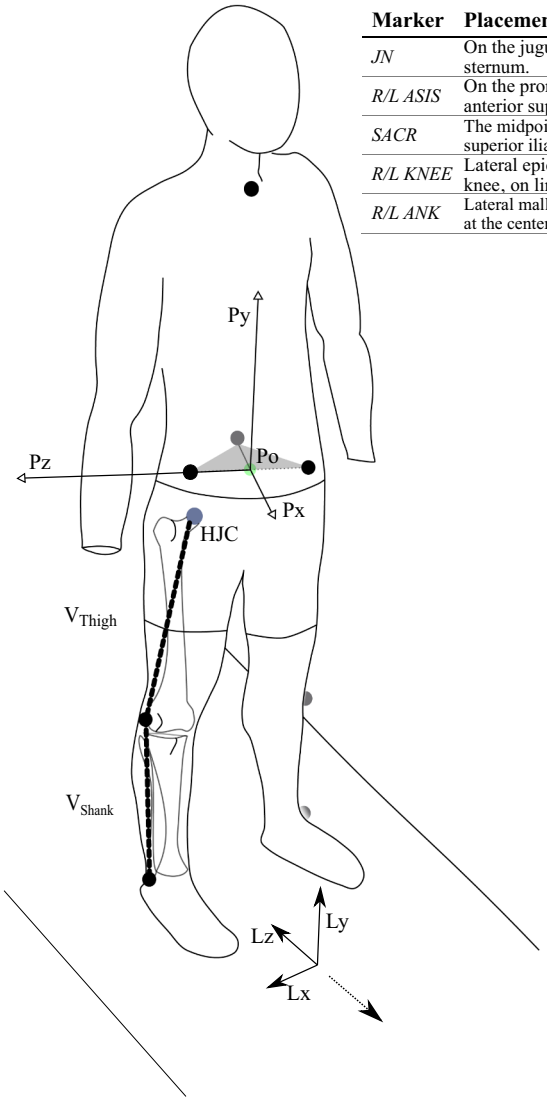

Fig. 2. Eight marker model (8MM) description and marker placement. V_{Thigh} is a vector from HJC to the knee, V_{Shank} is a vector from the knee to ankle. The origin of the pelvis (Po) was taken as the midpoint of R/L ASIS markers. HJC = hip joint center; L = lab axis; P = pelvis axis.

least 6 minute habituation to walking on the treadmill environment, self-selected, fixed comfortable walking speed was set. Participants then carried out 1 min of comfortable walking. Processed marker data from the full trial was then used for analysis.

2.4. Description of model

HBM was used to provide reference 'gold standard' gait data, against which the 8MM outcomes were compared. While HBM is not as widely implemented in clinical practice as other models, it is reliable in its biomechanical outcomes (Al-Amri et al., 2017; Falisse et al., 2018; van den Bogert et al., 2013), used in clinical practice (Booth et al., 2019; Gagliardi et al., 2018; Liu et al., 2017; Oudenhoven et al., 2019; Richards et al., 2017) and in validation studies (Karatsidis et al., 2018). Eight markers from the motion capture data were selected as input to the reduced marker model (Fig. 2): the jugular notch, the two most anterior margins of the iliac spines, the sacrum, i.e. the midpoint of the two posterior iliac spines, and bilaterally on the lateral epicondyle of the knee and lateral prominence of the lateral malleolus of the ankle. Data was reprocessed in D-Flow, with pelvis, hip and knee angles calculated in real-time using a custom script function, operating at ~300 Hz. A simplified version of the custom script, defining calculation steps of the model is available in supplementary material.

For 8MM, the pelvis was considered the root segment with six degrees of freedom relative to the world. Spatial rotations of the pelvis were calculated as reported following international recommendations (Wu and Cavanagh, 1995). The pelvis z-axis (Pz) was defined as the vector from RASIS to LASIS. The Py was defined as perpendicular to the Pz and a temporal Px, which was a vector from sacrum to the origin of the pelvis. The Px was then taken as perpendicular to the Py and Pz. Angles were reported relative to global laboratory coordinate system such that rotations about Pz (pelvic tilt), Px (pelvic obliquity), and Py (pelvic rotation) following Grood and Suntay (1983).

The hip joint center (HJC) was defined following the method reported by Harrington et al. (2007). Hip flexion-extension was calculated by projecting the vector from the HJC to knee (V_{Thigh}) on the pelvis sagittal plane. Knee flexion was taken as the inner product of V_{Thigh} and V_{Shank} (the vector from the knee to ankle).

2.5. Data analysis

Initial contact and toe-off events were based on marker data following Zeni et al. (2008), using only ankle markers for the foot in 8MM. Gait data were time-normalized to gait cycle (0–100%). Only the right

Table 2
Comparison of mean values and standard deviations for clinically relevant outcome parameters (CROPs) between the two models.

	TD (n = 41)			CP (n = 25)		
CROP Mean (SD)	HBM	8MM	*P*	HBM	8MM	*P*
Mean pelvic tilt (°)	10.9 (5.9)	10.9 (5.8)	0.880	17.0 (5.1)	17.9 (5.3)	0.009[a]
ROM pelvic tilt (°)	4.6 (1.0)	4.8 (1.0)	0.167	7.0 (2.4)	6.4 (2.1)	0.010[a]
ROM pelvic obliquity (°)	11.2 (2.4)	8.5 (2.1)	< 0.001[a]	9.6 (3.3)	6.6 (2.3)	< 0.001[a]
ROM pelvic rotation (°)	11.1 (3.3)	10.8 (3.0)	0.315	13.6 (3.8)	13.2 (3.9)	[NP] 0.427
Minimal hip flexion (°)	−6.3 (7.4)	−6.9 (6.5)	[NP] 0.026[a]	5.8 (9.3)	5.7 (8.6)	0.852
ROM hip flexion (°)	43.1 (5.5)	40.9 (5.0)	< 0.001[a]	39.9 (7.5)	37.1 (7.1)	< 0.001[a]
ROM knee flexion (°)	65.4 (6.0)	52.4 (5.2)	< 0.001[a]	47.7 (11.9)	41.1 (9.5)	[NP] < 0.001[a]
Minimal knee flexion at IC (°)	3.1 (5.4)	12.0 (3.1)	[NP] < 0.001[a]	25.8 (11.2)	27.0 (9.4)	0.076
Maximal knee flexion (°)	66.0 (5.2)	63.4 (4.5)	< 0.001[a]	61.6 (6.1)	59.2 (6.8)	0.001[a]
Step length (m)	0.50 (0.09)	0.50 (0.08)	0.740	0.37 (0.09)	0.36 (0.09)	0.798
Step width (m)	0.22 (0.03)	0.22 (0.03)	[NP] < 0.001[a]	0.27 (0.06)	0.26 (0.06)	< 0.001[a]
Cadence (steps/min)	116 (16)	119 (13)	[NP] 0.176	100 (13)	104 (21)	[NP] 0.412

[a] Indicates significant difference of CROP between models (α = 0.05). CP = cerebral palsy; IC = initial contact; TD = typically developing; HBM = human body model; 8MM = eight marker model; ROM = range of motion; NP = non-parametric test carried out.

leg was used in analysis. A set of clinically relevant outcome parameters (CROPs, Table 2) were extracted from each stride of each individual before averaging to obtain a group mean.

2.6. Usability

To assess the usability of the application in clinical practice, the application was distributed to five specialist rehabilitation centers across the Netherlands, using the GRAIL regularly in clinical practice. The application featured simple documentation on use and therapists were requested to use it as part of standard clinical care in any patient undergoing gait retraining. There was a trial period of one month, in which therapists were free to use the application as desired. At the end of the trial period, users were asked to fill out the System Usability Scale (SUS) (Brooke, 1995). This scale has been extensively tested for usability of technology, and shown to be valid, reliable and sensitive (Bangor et al., 2008). Ten statements are scored using a five-point Likert scale. The SUS provides a point estimate of percentage usability. Scores of above 85 are considered excellent, while scores below 50 indicate unacceptably low levels of usability (Bangor et al., 2008). Scores between 50 and 70 may be considered marginal/OK (Bangor et al., 2008). While not as established as the SUS, an additional set of short questions, specific to VR technology in rehabilitation (USEQ) (Gil-Gómez et al., 2017) was also included. The scale ranges from 6 (lowest satisfaction) to 30 (highest user satisfaction) (Gil-Gómez et al., 2017).

2.7. Statistics

To identify differences in the model outcomes for kinematic variables across the gait cycle, statistical parametric mapping (SPM) paired *t*-tests were carried out (α = 0.05) for each group separately. For each SPM t-test, a statistical parametric map (SPM{t}) was created by calculating the conventional univariate t-statistic at each point of the gait curve. If at any time, an SPM{t} crossed the critical threshold, a cluster was created, indicating a significant difference between two kinematics curves at a specific point of the gait cycle (Nieuwenhuys et al., 2017; Pataky, 2010). As errors in 3D gait analysis of between 3 and 5° are commonly considered acceptable (McGinley et al., 2009), a significant difference was interpreted as clinically relevant if the mean waveforms were at least 5° removed (RMSE) from each other, on average, within the areas of significance as indicated by the SPM analysis. All SPM analyses were performed using open-source SPM1d code (version M.0.4.5; www.spm1D.org) in MATLAB (version 2017a, MathWorks Inc., USA). To further quantify differences between models, within-subject RMSE values between average kinematic curves over the gait cycle were calculated for each subject and averaged over each group.

The set of CROPs extracted from the TD and CP group were tested for normality using Shapiro-Wilk tests. If normality was not violated, difference between groups were tested using a paired t-test, otherwise a Wilcoxon signed-rank test was used (α = 0.05).

3. Results

3.1. Difference in model outcomes

In both the TD and CP group, outcome of the two models were comparable, with significant but small differences identified, presented in Figs. 3 and 4.

For the TD group, all kinematic output - except for pelvis tilt - differed significantly within at least one area of the gait cycle ($p < 0.05$, Fig. 3). For the pelvis, the areas in which SPM indicated significant difference, the RSME between models was small (< 3°). For hip extension, three significant clusters (0–1, 20–23, 70–100% of the gait cycle) exceeded the critical threshold where 8MM showed considerably less hip flexion than HBM (RMSE = 5–6°). The knee angle also showed significant difference between models. Two significant clusters (0–66, 95–100% of the gait cycle) exceeded the critical threshold where 8MM showed considerably less knee extension compared to HBM (RMSE = 6.8, 8.1°, respectively). One further supra-threshold cluster (74–91%) exceeded the critical threshold where 8MM showed less knee flexion than HBM (RMSE = 4.6°). Within-subject RMSE between the models, across the gait cycle was generally small (< 4°) for all kinematic outcomes. However, knee extension was the exception to this, where considerably reduced maximal extension and flexion was found, resulting in larger errors between models (RMSE = 5.6° (SD 3.2°)).

The results in the CP group data showed a similar trend. All parameters differed significantly within at least one phase of the gait cycle ($P < 0.05$, Fig. 4). For all clusters SPM indicated a significant difference, the RSME was small (< 5°). An exception to this was knee extension, where one supra-threshold cluster (44–53%) exceeded the critical threshold where 8MM showed less knee extension than HBM (RMSE = 5.2°). Within-subject RMSE between the models, across the gait cycle was smaller than that of the TD group and < 4° error for all kinematic outcomes.

CROPs for each group and model outcome are presented in Table 2. In the TD group a significant, yet slight increase in range of motion of pelvis obliquity was found in 8MM ($P < 0.001$). The CP group showed significant yet small reductions in ROM pelvis tilt and mean pelvis tilt compared to HBM ($P = 0.009$ and 0.010, respectively). ROM at the hip was significantly smaller in 8MM for both TD and CP groups ($P < 0.001$). The largest differences were found at the knee, where a considerable (> 10°) decrease in range of motion was observed in the

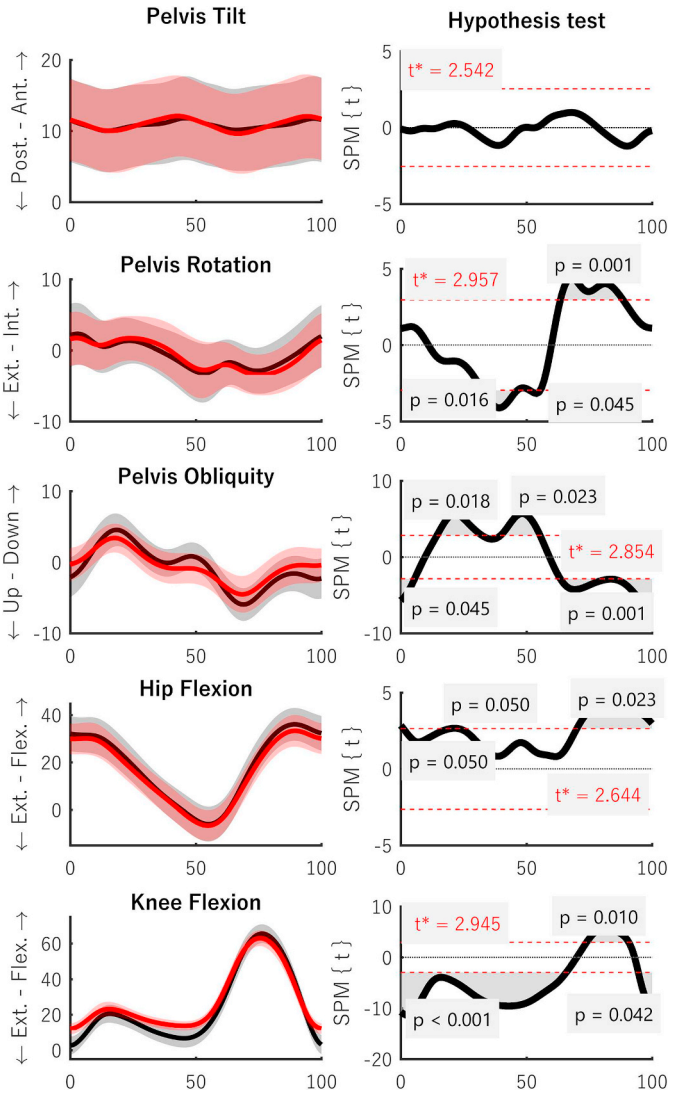

Fig. 3. Comparison of gait kinematic outcomes for TD group. Left graphs show the group average kinematics and 1SD of HBM (black) and 8MM (red). Right graph shows results of SPM paired *t*-test statistic as a function of the gait cycle. Clusters in which the critical threshold (t*, red dashed line) was exceeded are shaded in grey with level of significance, this indicates significant difference between models at this area of the gait cycle. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

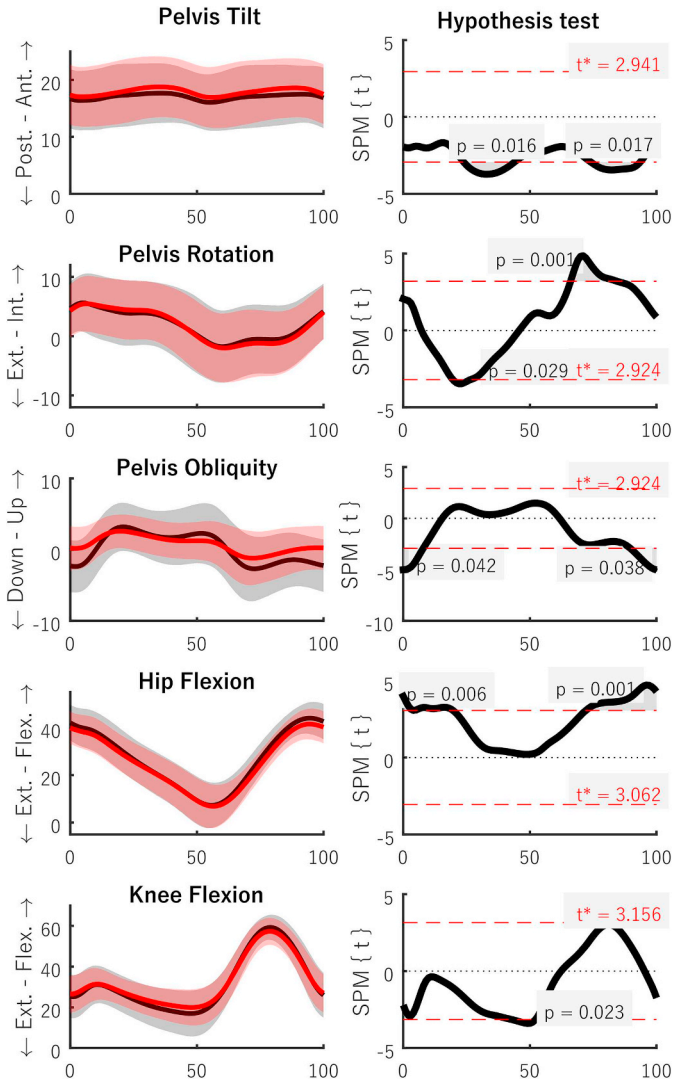

Fig. 4. Comparison of gait kinematic outcomes for CP group. Left graphs show the group average kinematics and 1SD of HBM (black) and 8MM (red). Right graph shows results of SPM paired t-test statistic as a function of the gait cycle. Clusters in which the critical threshold (t*, red dashed line) was exceeded are shaded in grey with level of significance, this indicates significant difference between models at this area of the gait cycle. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

8MM ($P < 0.001$). The difference between models at the knee joint was larger in the TD group than in the CP group. This was also reflected in the reduction of minimal knee extension at initial contact for the TD group ($P < 0.001$), however, this was not significant for the CP group ($P = 0.076$).

Spatiotemporal outcomes were comparable. No difference in step length was identified, however, a significant, yet negligible reduction in step width was identified in both groups ($P < 0.001$).

3.2. Usability

In qualitative assessment of the application, all responders (n = 6) had previous experience and training on the system and the general interface. The users regularly implemented the GRAIL for clinical gait analysis and training in a range of patient populations. The mean SUS score for all responders was 68 (SD 13) (range: 50–85). The mean USEQ scores was 24 (SD 3) (range: 20–30).

4. Discussion

In this study we presented a novel application to provide immersive visual biofeedback on clinically relevant aspects of gait. The application is based on a simplified biomechanics model, requiring only 8 markers and consequently a short set up time. The model outputs kinematic and spatiotemporal data in real-time, which may be used to provide biofeedback with the goal of achieving an improved gait pattern in patient populations undergoing gait re-training. In addition, with placement of eight markers on anatomical landmarks, it is possible to visualise trunk and lower limb movement with an avatar and augmented biofeedback.

The 8MM was found to output gait kinematics in real-time that were comparable to a 'gold standard' biomechanical model (HBM). The differences were mostly small and can be considered to be within acceptable range (< 5°) (McGinley et al., 2009). While statistically significant differences were identified, in both SPM analysis of waveforms and CROPs, they were small (< 3°).

The lower ROM of pelvis obliquity in 8MM is an interesting finding.

This may be partially explained by the model definitions. In 8MM, the pelvis is a free segment, calculations are carried out iteratively and updated at each frame. HBM utilizes global optimization of all marker positioning that introduces marker error between the measurement and the (global) model (van den Bogert et al., 2013). This may have the effect of a slight increase in the pelvis ROM as the pelvis orientation is now influenced by other segments.

Further discrepancies between the model can be found at the hip and knee, where reduced range of motion was observed in 8MM. While differences in hip ROM between the models can be considered minor ($< 3°$), differences at the knee were starker. This was most evident in the TD group, where ROM of knee flexion was considerably smaller in 8MM ($> 10°$). This primarily relates to a reduction in maximal extension at the knee. This can be explained by the simplified method of knee angle calculation; the inner product of V_{Thigh} and V_{Shank}. The origin of the thigh is considered the HJC, projecting laterally to the knee, while the shank vector was based on the lateral positioning of the knee and ankle marker. As such, even when the subject is standing straight, full knee extension may not be observed. In contrast, HBM uses subject specific calibration of the model, with accurate joint centers and segment angles calculated relative to these reference frames. A static calibration step may be implemented in the 8MM to increase robustness, however, this would add to set up time. An additional solution to improve knee extension estimation for 8MM may be to include only the sagittal component of the vectors. However, in pathological gait this may act to reduce reported knee flexion. As the presented tool is intended for use in individuals with walking pathology, this method of knee joint calculation was upheld. The importance of differences in the model output is further negated by the intended use of the application. While it is important to provide accurate biofeedback, the goal of the application is to challenge improved performance relative to an individual's baseline walking within the session. As such, during each session, the patient's comfortable walking is recorded and used to set targets for improvement, e.g. improve knee extension at initial contact by 5°. Therefore, the subject acts as an internal control. So long as data is of reasonable accuracy and similarity to 'gold standard' output, it can be considered valid for intended use, as demonstrated in this analysis. Inter-session reliability is an important factor that could not be assessed, due to the use of retrospective data from individual gait sessions to validate the model. Inter-session reliability would be a valuable direction of further investigation.

Incorporating VR and biofeedback into rehabilitation is important in maintaining engagement in prolonged training interventions, however, it must be clinically usable and based on sound therapeutic reasoning. Therapy sessions are often limited in time by financial pressure. With the tool presented in this study, no subject calibration is required, and marker placement set up time is reduced to < 5 min, compared to that of roughly 30 min for typical marker set-up of model in gait analysis. Thus, the time available for training is maximized, increasing the potential intensity of intervention. Implementation of the 8MM, however, is a trade-off between robustness and simplicity. It is therefore more susceptible to poor marker tracking and discrepancies in feedback provided and true angles, especially towards full knee extension. As such, if implemented as a gait training tool in clinical practice, the patient may undergo a typical gait analysis, with HBM, at the start and end of the programme. This data would be more robust and informative for tracking progress of the patient. The instability of the visual biofeedback as a result of marker drop out was highlighted as a negative in the SUS responses. Given these limitations, the application was found to score well on the USEQ (24 (3)), with responses indicating this would be a valuable addition to gait rehabilitation (Gil-Gómez et al., 2017). The responses from the SUS were more mixed, yet promising. Overall it can be considered of 'high marginal acceptability' (mean 68 (SD 13)) (Bangor et al., 2008). In interpretation of the SUS, average score (at the 50th percentile) is 68 (Bangor et al., 2008). This result implies that the application is acceptable for passive users but improvements are required for users to actively promote uptake of the application. Further work to simplify the application interface and stability of the model is required to maximise the usability.

The present study was subject to a number of limitations. The limited number of responders allows for only suggestive evidence among a specific group of specialist therapeutic centers and may not be applicable to general therapeutic practice. The sample size was limited by the number of centers with the available technology to trial this application as a rehabilitation tool. The presented set-up was developed on a commercially available, high cost, advanced treadmill system. The cost of such a system presents a significant barrier to the wider uptake of its use in clinical practice. However, the method described may not be restricted to this system. 8MM uses simple mathematical calculations to define clinically important aspects of gait. In theory, this may be used with a simple treadmill; any form of motion capture streamed through a suitable software package; and connected to a screen for visualisation. Such equipment is more commonly available in rehabilitation centers. Further work should be undertaken to establish if this form of biofeedback enhanced gait training can result in gait improvements in the long term. This should be accompanied by cost effectiveness analysis to justify the use of such systems. Developments in marker-less motion capture technology may result in lower cost alternatives for providing feedback on gait. In addition, biofeedback should not be limited to the laboratory environment. Inertial measurement units (IMUs) technology may provide solution to this (Gailey et al., 2017; Karatsidis et al., 2018; Washabaugh et al., 2017). Integration of multiple forms of biofeedback and VR, both in the lab and at home, may result in the best treatment outcomes by providing variable, simulating and task specific practice.

5. Conclusions

This study investigated the validity and usability of 8MM to facilitate avatar-based, biofeedback-enhanced gait training. It was found that 8MM could provide acceptable accuracy of gait kinematics and spatiotemporal parameters in real-time, when compared to a robust biomechanical model, for the application in gait training. However, due to simplification of calculations, knee extension was significantly diminished in the 8MM, which should be considered when implementing the model. Further investigation is required to establish the effectiveness of biofeedback enhanced gait re-training in improving functional outcomes relating to walking. If it is found to be effective, low-cost, widely accessible alternatives may be explored.

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Declaration of competing interest

We wish to draw the attention of the Editor to the following facts which may be considered as potential conflicts of interest to this work. ATCB is employed by Motek Medical B.V. in a position fully funded by the PACE ITN, under the European Union's Horizon 2020 research and innovation programme, Marie Skłodowska-Curie grant agreement number 642961. FS is employed by Motek Medical B.V. No other authors have any conflict of interest to declare. All research direction and integrity is supervised by VU University Medical Center.

Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.clinbiomech.2019.08.013.

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