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Visual kinematic feedback enhances the execution of a novel knee flexion gait pattern in children and adolescents

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ABSTRACT

Background: Altered knee motion is one of the most common gait deviations in pediatric populations with gait disorders. The potential for pediatric gait retraining using visual feedback based on knee kinematic patterns is under-explored.
Research question: This study investigated whether pediatric participants could successfully modify knee flexion patterns in response to a visual kinematic feedback system (VKFS).
Methods: Knee flexion angles from twelve typically developing children and adolescents (6 M, 6 F; 11.9 ± 2.7 years) were calculated using wearable inertial measurement units. Participants were tested while walking on a treadmill using pattern based visual feedback (FB). Four novel target patterns which amplified or attenuated swing phase peak knee flexion were tested. No feedback (NFB) tests assessed the participant's ability to independently reproduce the patterns. Mean absolute cycle error (MACE) and magnitude of peak knee flexion error (PK) were calculated during the last 10 strides of FB and NFB trials. Pre-exposure reference values (R) were also calculated.
Results and significance: PK-FB was significantly smaller ($p < 0.05$) than PK-R for all targets. Average values for PK-NFB were higher than for PK-FB, although PK-NFB remained significantly lower than PK-R for two targets. Contrary to one of the study's hypotheses, MACE-FB and MACE-NFB were larger than MACE-R.

The study provided evidence that pediatric participants were able to modify peak knee flexion during gait in the sense targeted by the VKFS. Analysis suggested that MACE increases were explained by increases in gait cycle deviation outside of the changed region.

1. Introduction

The 2012 Americans with Disabilities Report estimated that around 580,000 children in the United States under the age of 15 experience difficulty with walking or running [1]. Such disability impacts functional mobility and can affect social development and engagement [2]. Pathological gait patterns are also associated with long term development of musculoskeletal injuries such as osteoarthritis [3,4], and these factors are particularly critical when onset is at birth or during childhood. Cerebral palsy (CP) [5] and, with a much lower incidence,

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spinal cord injury (SCI) [6,7] are two diagnoses that are frequently associated with such disability.

Altered sagittal knee motion is one of the most common deviations associated with reduced gait function [8,9]. Nieuwenhuys [10] and Krawetz and Nance [11] reported a variety of temporal and spatial deviations in knee flexion patterns associated with CP and SCI. These deviations from normative gait have severe consequences for gait function. For example, reduced knee extension in terminal swing leads to shortened stride length which results in a decrease in velocity, whereas a reduced peak knee flexion during mid swing is associated with loss of foot clearance and can increase the risk of falls [8,12].

Both surgical and therapeutic approaches have been developed to address gait deviations. Most therapeutic approaches have employed repetitive training provided by treadmill walking, complemented by either manual or robotic mechanical guidance of limb segments towards normalized temporal and spatial gait parameters. Using such methods, reduced deviations in knee kinematics and improvements in functional measures have been demonstrated in ambulatory CP [13] and SCI [14] patients. These methods are inherently resource intensive; for instance, both robotic training equipment and commitments of manual therapeutic training time are expensive. Biofeedback retraining paradigms are based on the patient's active interpretation and response to feedback cues of gait pattern deviations. They require sensing, processing, and output (e.g. visual, haptic, or auditory) technologies that tend to be less expensive than the hardware required to generate and control assistive forces. Whereas active limb assistance will continue to be a necessity in many patients, there is evidence that within both SCI and CP populations there exist subsets of patients that exhibit pathological kinematics despite normal, or near normal, voluntary joint ranges of motion. Schließmann [15] demonstrated that five ambulatory adults with incomplete SCI and predominantly sensory deficits could respond appropriately and retrain to unilateral feedback on knee joint kinematics. In the CP population, a recent study [16] showed that a pattern of increased knee flexion at initial contact was significantly correlated with a deficit in selective motor control for the lower extremity. In contrast with other patterns that were correlated with structural findings such as muscle contractures, this learned dysfunctional motor pattern may be addressable by feedback-based gait retraining.

During biofeedback-based gait training, real-time visual [17], auditory [18], or haptic [19] information augments native sensory and visual feedback. Successful use of visual feedback relies on the sensory integration of visual signals to modify specific motor behaviors [20], and several studies have reported that adults and pediatrics differ in the speed and manner of processing and integrating visual, vestibular, and somatosensory information [21–23]. Moreover, successfully responding to feedback cues depends on functional task complexity [24], which gauges the level of difficulty of the task in reference to the cognitive capacities of the user. On the one hand, while varying success has been reported in the use of biofeedback modalities that focus on one specific target parameter resulting from the gait cycle [25–27], these techniques do not target deviations in other parts of the cycle, and might therefore induce new deviations to the learned pattern. On the other hand, pattern based training with whole gait cycle feedback offers the advantage of specificity of guidance toward a desired target pattern, but the task of appropriately responding represents higher functional task complexity. The ability to appropriately respond to visual feedback based on the entire gait cycle has been demonstrated in adults [15,17], but no prior evaluations of the performance of pediatric participants with whole gait cycle feedback were found in our literature search.

The purpose of the current study was therefore to evaluate the degree to which pediatric participants could successfully modify their gait patterns in response to whole gait cycle feedback provided by a visual kinematic feedback system (VKFS). This system uses low cost wearable hardware that is compatible with treadmill training, and provides feedback on the trajectory of knee flexion angle relative to a prescribed target pattern. We chose to perform this feasibility study in typically developing children without disabilities; this focused the study on the baseline cognitive and motoric compatibility of the modality with children in our age group and excluded variable effects of pathology that arise in target clinical populations. Four novel target patterns were studied in a repeated measures experiment, and errors in peak knee flexion angle and mean absolute error between the target and the observed gait cycles were measured. Our primary hypotheses were (1) that participants would respond to VKFS training with modified gait patterns that were closer than their baseline gait patterns to the targeted patterns, and (2) that participants would be able to maintain gait modifications after removal of feedback. We also descriptively compared between study results from the four target patterns.

2. Methods

2.1. Participants

This study included typically developing pediatric participants meeting the following inclusion criteria: 7–15 years of age, able to understand spoken English, and able to walk without difficulty on a treadmill. Exclusion criteria included: significant injury that interfered with the ability to walk, significant recent surgery, and known increased risk of stroke or heart attack.

All research procedures were approved by the institutional human subjects research review board.

2.2. VKFS

Each participant's dominant limb was assigned based on their response to the Waterloo Footedness Questionnaire [28]. Dominant limb orientation and acceleration data were collected using Inertia Measurement Units (IMU) (Xsens MTw, Enschede, The Netherlands) with a sampling frequency of 50 Hz. Two sensors were attached to the leg: one sensor was placed horizontally on the anterior aspect of the thigh, and a second sensor was placed on the posterior aspect of the shank with the long axis of the enclosure parallel to the thigh sensor. A third sensor was taped to the heel of the shoe with the long axis of the enclosure vertical during standing. To calculate knee flexion angles, formulate and display the feedback, and record experimental data, the VFKS was implemented using the MTw Devkit (Xsens) application programming interface and MATLAB (MathWorks Inc., Natick, MA). Gait cycle triggering was performed by detecting periods of 0.1 s representing the initial quiet period of foot IMU acceleration associated with heel contact and the initiation of stance phase. The display was updated after approximately 25% of the gait period had elapsed to reflect the previous stride, and included three different types of information (Fig. 1). The primary feedback display was a 180-degree dial for which the needle moved in proportion to the sum across the stride of squared knee flexion pattern angle error. Two assistive indicators were provided: a vertical indicator displayed the signed error between the actual and the target maximum knee flexion angles, and a horizontal indicator showed the signed error between the actual and the target timing of maximum knee flexion.

2.3. Target gait patterns

A reference gait cycle (RGC) was calculated offline for each participant by averaging 20 strides collected during their baseline treadmill walking. All analyzed strides were normalized in time (0–100%, interval = 1%) by spline interpolation of the raw data between consecutive gait cycle triggers. To calculate the target gait cycles (TGCs), the RGC curve during swing was modified between the two minima before and after maximum knee flexion. The side of the curve with the lower minimum was amplified and attenuated by 20 and 40%, the other side of the curve was scaled accordingly resulting in four different TGCs

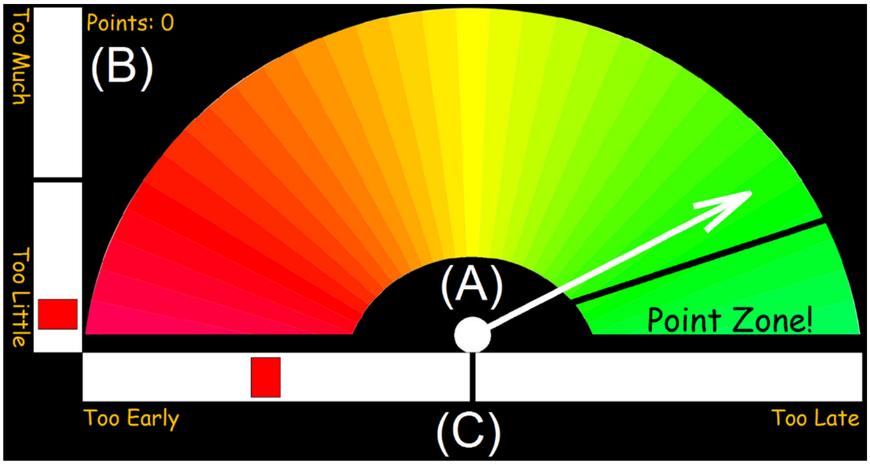

Fig. 1. Visual kinematic feedback system (VKFS) display showed (A) primary feedback needle for gait pattern deviation. Large errors were indicated by movement of the needle to the red area; full scale deflection of the needle (horizontal pointing left) corresponded to a root mean square error across the cycle of 24.4°. A 'horizontal pointing right' needle position signified zero error, and a point count was incremented each time the needle indication was in a 'point zone' that corresponded to a root mean square error across the cycle no higher than 7.7°. Assistive displays provide feedback on (B) magnitude and (C) timing of peak knee flexion.

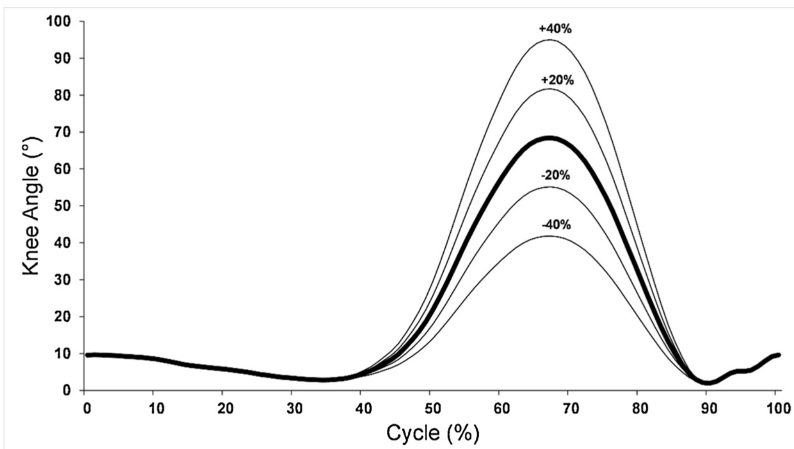

Fig. 2. Typical knee flexion curves for feedback testing. Bold curve indicates the average normal walking curve (RGC) and other curves indicate deviated target patterns (TGC).

(TGC_{+40}, TGC_{+20}, TGC_{-20}, TGC_{-40}) (Fig. 2).

2.4. Testing and protocol

The knee angle during standing was measured by goniometer, and an offset was applied to the IMU-calculated flexion angle to match this value.

Testing commenced with familiarization to walking on the treadmill while selecting a preferred self-selected speed which was used for all testing. Treadmill speed was adjusted by the examiner according to participant preference based on the instruction to select a comfortable, sustainable walking speed. Testing proceeded when familiarization was judged to have occurred based on visual assessment of gait and verbal confirmation of readiness.

Baseline walking data were collected during one minute of treadmill walking. The order of presentation for TGCs was randomized, and within each TGC two different conditions were tested in fixed order: Feedback (FB), and No Feedback (NFB). For FB, participants used VKFS cueing while walking on the treadmill for 3 min. Before testing the first FB condition, participants were introduced to the feedback display and the function of each indicator was explained (Fig. 1). Using age-appropriate lay language, participants were told that to gain points in the feedback tasks they would need to increase swing phase knee flexion for two targets and decrease it for two targets; they were not however told which of the targets was active during the trials. For NFB trials participants were instructed to try to maintain the modified pattern during 1 min of treadmill walking. Between testing of consecutive TGC's, participants rested for 3 min and were then asked to perform natural walking on the treadmill for 1 min.

2.5. Experimental measures

Two measures quantifying correspondence between observed (α_O) and target (α_T) knee angle trajectories were calculated, with values of α_T corresponding to the target cycles (TGC) defined above.

For each individual, *peak knee flexion error ('PK')* was calculated as the mean across m analyzed strides of the absolute errors between maximum peak of target and observed knee flexion angles:

$$PK = \frac{\sum_{i=1}^{m} |\max(\alpha_{O,i}) - \max(\alpha_{T,i})|}{m}$$

Mean Absolute Cycle Error ('MACE') was calculated as the mean across m analyzed strides of the mean absolute error between the 101 samples across the gait cycle of α_O and α_T:

$$MACE = \frac{\sum_{i=1}^{m} \sum_{j=1}^{101} |\alpha_{O,ij} - \alpha_{T,ij}|}{101m}$$

In order to quantify responses primarily within and outside of the regions of the targets that were manipulated, we also calculated MACE for subregions of the cycle. These were chosen to approximate stance and swing phases - $MACE_{SWING}$ and $MACE_{STANCE}$ respectively. The tracking system was not designed to detect toe off (foot off) events; therefore, the periods were approximated by assuming the initial 60 percent of the gait cycle corresponded with stance, and the final 40 percent corresponded with swing. Previous studies indicate that stance phase duration is typically around 60 percent of the total gait cycle for walking in both children [29] and adults [30].

For each target trial, a single value for each measure was thereby calculated for each subject in three conditions: *Reference condition (R):* the RGC values were used as α_O; this produced reference measures representing variance in the RGC pattern from the TGC as well as stride-to-stride variance.

FB and NFB conditions: α_O values were from the last m = 10 strides observed during FB and NFB trials, respectively, for each TGC.

2.6. Statistical analysis

Significance level of 0.05 was used for all statistical testing. For all TGCs in FB and NFB conditions, paired sample t-tests were used to test for significant differences between each of the above measures and the respective reference measure ($MACE_R$, PK_R). Separate one way repeated measures ANOVAs were also used to test the effect of TGC on MACE and PK during FB and NFB. If a significant main effect was found, Bonferroni post hoc comparisons were used to determine pair-wise differences. Descriptive statistics (mean ± standard deviation) were calculated for each measure.

3. Results

3.1. Enrollment and consistency of baseline walking

Twelve typically developing children and teenagers (6 M, 6 F; 12.4 ± 2.7 yrs., range 8.1–15.6 yrs.) participated in the study. Baseline gait variability, assessed by calculating coefficients of variation for stride duration for each participant was 3.5 ± 1.0%.

3.2. Peak knee flexion error

PK_{FB} was significantly smaller than PK_R for all targets (Fig. 3, TGC_{+40}: $p < 0.001$; TGC_{+20}: $p = 0.022$; TGC_{-20}: $p = 0.042$; TGC_{-40}: $p < 0.001$). Average values for PK_{NFB} were higher than for PK_{FB}, although PK_{NFB} remained significantly lower than PK_R for two target cycles (TGC_{+40}: $p = 0.001$, TGC_{-40}: $p = 0.001$).

3.3. Mean Angle Cycle Error

$MACE_{FB}$ was significantly higher than $MACE_R$ for target angles TGC_{+20} ($p = 0.029$) and TGC_{-20} ($p = 0.043$). No significant differences between $MACE_{FB}$ and $MACE_R$ were observed for TGC_{+40} ($p = 0.227$) and TGC_{-40} ($p = 0.239$) (Fig. 3). Average $MACE_{NFB}$ was higher than $MACE_{FB}$ for all target angles, and $MACE_{NFB}$ was significantly higher than $MACE_R$ for all target angles (TGC_{+40}: $p = 0.025$; TGC_{+20}: $p = 0.030$; TGC_{-20}: $p = 0.014$; TGC_{-40}: $p = 0.018$).

$MACE_{STANCE,FB}$ was significantly higher than $MACE_{STANCE,R}$ for all target angles (TGC_{+40}: $p = 0.015$; TGC_{+20}: $p < 0.001$; TGC_{-20}: $p = 0.01$; TGC_{-40}: $p = 0.001$) (Fig. 3). Average $MACE_{STANCE,NFB}$ was lower than $MACE_{STANCE,FB}$ for target angles TGC_{-20} and TGC_{+20}, and higher for target angles TGC_{-40} and TGC_{+40}. $MACE_{STANCE,NFB}$ was significantly higher than $MACE_{STANCE,R}$ for all target angles (TGC_{+40}: $p = 0.006$; TGC_{+20}: $p = 0.009$; TGC_{-20}: $p = 0.003$; TGC_{-40}: $p < 0.001$).

$MACE_{SWING,FB}$ was significantly lower than $MACE_{SWING,R}$ for target angles TGC_{-40} ($p = 0.001$). No significant differences between $MACE_{SWING,FB}$ and $MACE_{SWING,R}$ were observed for TGC_{+20} ($p = 0.231$), TGC_{-20} ($p = 0.520$), and TGC_{-40} ($p = 0.281$) (Fig. 3). Average $MACE_{SWING,NFB}$ was higher than $MACE_{SWING,FB}$ for all target angles. No significant differences between $MACE_{SWING,NFB}$ and $MACE_{SWING,R}$ were observed (TGC_{+40}: $p = 0.808$; TGC_{+20}: $p = 0.231$; TGC_{-20}: $p = 0.072$; TGC_{-40}: $p = 0.734$).

3.4. Comparisons of results between target cycles

No significant differences were observed between TGCs for PK and MACE. However, for both MACE and PK during FB, TGC_{+20} and TGC_{+40} tended lower than the targets that required decreased knee flexion (TGC_{-20} and TGC_{-40}). During NFB, TGC_{+20} tended to present the lowest MACE and PK.

4. Discussion

This study investigated whether typically developing pediatric participants responded appropriately to visual kinematic feedback across four targeted gait deviation patterns. PK and MACE were used to describe the similarity of the target and observed knee flexion patterns with feedback exposure, and while reproduction of the targeted patterns was attempted. Baseline gait variability was similar to values of 2.1 ± 0.1% and 3.3 ± 0.2% previously reported in over ground walking in 6–7 and 11–14 year old subjects [31], suggesting that participants were familiar with treadmill walking before commencement of targeting tasks.

During feedback exposure, peak knee flexion errors were reduced from those in the reference gait. Since the defining features of the target cycles were the changes in peak knee flexion, we interpret this as evidence that participants were able to interpret the visual feedback and respond with directionally appropriate gait adaptations. Quantitatively, the value of PK_{FB} averaged across the targets was 9.86 ± 3.16 degrees. In a pilot study in nine able bodied adults, Schließmann [15] reported lower tracking error levels with target patterns that were similar to those we studied. Several aspects of the error assessment and the feedback presentation cloud the comparison of these values to those in our study. Schließmann's values are pooled signed errors across participants, computed after a number of strides for which the pooled group performance dropped below a pre-defined performance criterion. Due to offsetting negative and positive deviations, averaging signed rather than absolute deviations produces lower mean errors; also, reporting data based on a performance criterion by definition selects for low error values within each trial. In our study, we observed that performance and attention varied within trials; we chose to report data from the final strides as objective and consistent measures of performance. We did not provide additional cueing to encourage engagement during this period, and our measures under-estimate task performance during highest engagement. Moreover, the error metric reported by Schließmann was the same as the one dimensional feedback signal given, a measure of average knee joint angle error during the swing phase. Lower errors may be expected with this direct relationship than in our assessment of peak flexion error in the presence of three simultaneous cues.

In contrast with the findings for PK, MACE increased from the reference condition with feedback exposure, with significant increases being found for two of the four TGCs. To understand the increases in mean absolute deviation despite reductions in peak error, we examined the results for sub-regions of the gait cycle corresponding to stance and swing phases (Fig. 3). We observed that $MACE_{STANCE}$ tended to increase with feedback, having qualitative similarity to MACE, whereas $MACE_{SWING}$ showed some evidence of reduction with feedback and resemblance to the PK findings. A frequently observed finding that explains this is illustrated in example data from a single participant (Fig. 4), wherein desired adaptations in the altered swing phase region persisted as generalized changes with increased or reduced knee flexion across the gait cycle. We examined the possibility that these changes in other phases were biomechanically required compensations to maintain balance or forward progression relative to the treadmill belt. However, in qualitative observations made during pre-study testing in four adults, we noted consistent targeting within a few degrees of absolute error across the gait cycle. We do not know of fundamental differentiating biomechanical factors that would prevent pediatric participants from responding similarly. Instead, based on our observations we believe

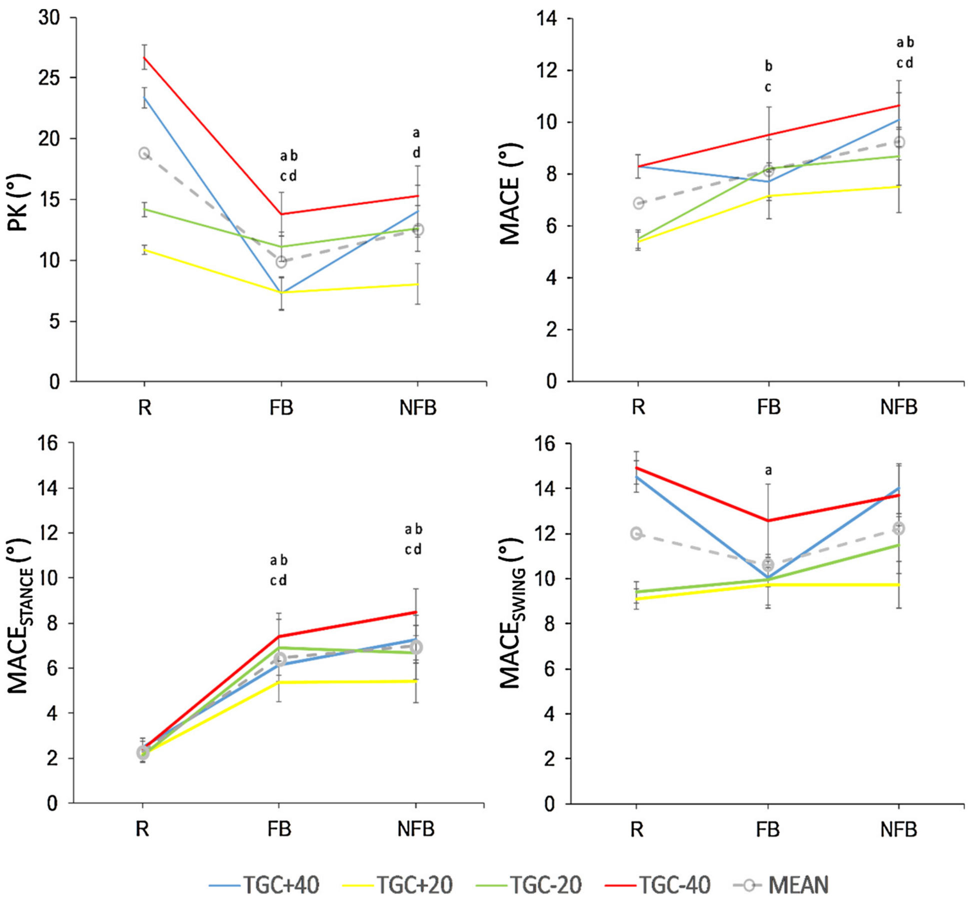

Fig. 3. PK, MACE, $MACE_{STANCE}$ and $MACE_{SWING}$ results for target gait cycle (TGC) across testing conditions R (Reference), FB (Feedback), and NFB (No Feedback). Error bars denote group standard errors; superscripts indicate significant differences from R for FB and NFB at different targets: (a) TGC_{+40}, (b) TGC_{-40}, (c) TGC_{+20}, (d) TGC_{-20}.

that the exposure time limited the participants' ability to consistently match the full target cycle. Participants tended to be focused on the assistive cues until late in the trial, and were still learning how to consistently score points based on the primary feedback. It has been suggested that motor learning of complex tasks is processed in three phases [32]: at first an initial representation of the movement is developed, secondly the movement is refined, and finally the movement is fully learned through repetitive practice. Our observation seemed consistent with transition between the first and second phases, with considerable variability in the time participants spent in each phase. Longer training times may have led to better results in respect to a lower absolute error and less variability. Also, previous work has highlighted the sensitivity of MACE to timing variations, showing that variability in the timing of peak knee flexion and in peak knee flexion magnitude have similar power to predict MACE variability [33]. This implies that small deviations in timing of swing phase knee flexion can lead to large contributions to MACE, even when patterns are qualitatively alike and excursion values are similar. This observation can further explain our findings on MACE. For instance, while as mentioned above PK and $MACE_{SWING}$ results showed similarity, there were also differences: in the TGC_{+20} and TGC_{-20} for the targets, PK error was reduced but $MACE_{SWING}$ was increased somewhat due to timing variability. This observation can also help to inform the choice of feedback signals for training. The primary feedback used in this study was driven by a time averaged absolute error metric similar to MACE. For training applications that seek primarily to influence joint excursions (active range of motion) high sensitivity of feedback to timing variability during practice may not be desirable. Our current development work is exploring the use of alternative signal similarity measures for feedback that are less sensitive to timing variability (for example, Dynamic Time Warping [34]). Furthermore, we performed a sub-analysis of performance stratifying to two sub groups above and below the median age (Table 1). Findings of higher performance in both PK and MACE in the older group illustrated developmentally-related differences in the ability to integrate and respond to feedback within our sample. This result together with the better performance seen in adults during pre-study testing underlines the importance of further investigations on the perceived workload of users during feedback, on the users' satisfaction with the feedback implementation, and – eventually by using eye-trackers – on the identification of the most relevant components of the visual feedback interface.

PK and MACE measures were also observed as participants attempted to reproduce the targeted gait patterns without feedback. PK_{NFB} increased relative to PK_{FB} for all targets, indicating that gait cycle tracking became less accurate without feedback. However, for two of the four targets the errors remained significantly lower than in the reference condition, and overall the participants showed evidence of being able to reproduce aspects of the trained gait cycle following the brief feedback exposure period. Distributed practice training paradigms incorporate training blocks with no feedback or reduced feedback and have been shown to promote the internalization of learned skills [35,36]. Our findings provide some evidence in support of future use of visual kinematic feedback training in such paradigms.

The four TGCs were selected to test performance across a range of conditions, and although contrasts between TGCs were not a primary research focus, we chose to conduct a descriptive analysis of differences between TGCs. This yielded the findings that PK_{FB} was higher for the two conditions in which targeted knee flexion was reduced (TGC_{-20} and TGC_{-40}), and that the highest $MACE_{FB}$ was observed for TGC_{-40}. This was consistent with findings of the study by Schließmann [15], wherein larger error values were recorded for 'reduced knee flexion' targets. These findings are consistent with the clearly more challenging nature of these tasks; as observed clinically, reduced knee flexion requires secondary motor adaptations such as ipsilateral hip abduction or

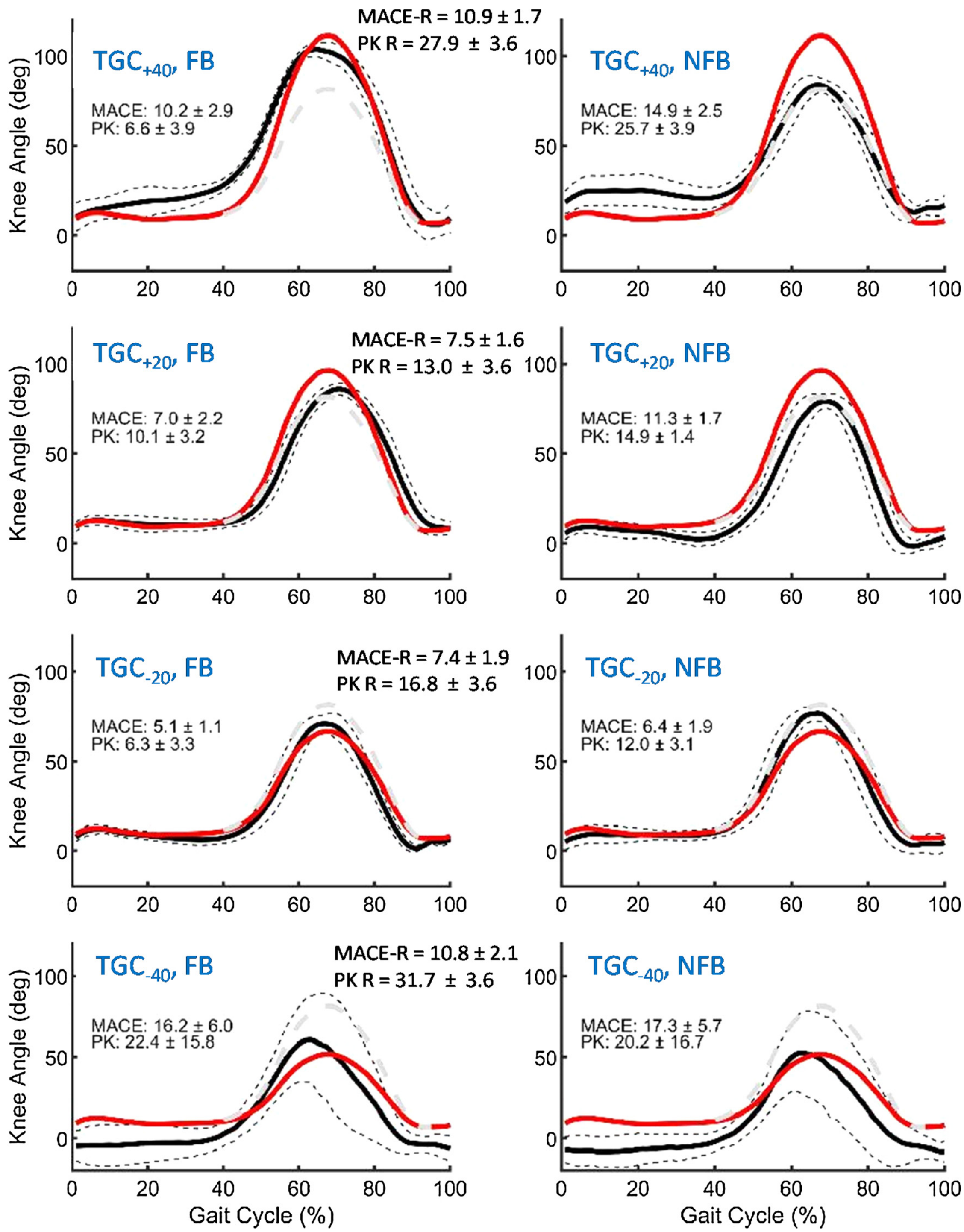

Fig. 4. Time normalized gait data for the analyzed period (final ten strides) of one participant (male, 14 yrs). Key - solid and dotted black: mean measured knee flexion pattern ± 1 S.D.; red: target pattern; dashed gray: mean baseline gait pattern for the participant. PK values and visual interpretation illustrate trends of (1) movement from reference condition toward target pattern with feedback; (2) maintenance of aspects of trained pattern without feedback, with reduced performance compared to condition with feedback. MACE values indicate (1) similar values or increases from reference condition with feedback; (2) increase of MACE on withdrawal of feedback. In TGC + 40 and TGC-40 conditions, increases or maintenance of MACE despite decreases in PK were explained by stance phase adaptations consistent with the change cued for the swing phase peak knee flexion (flexion and extension relative to baseline respectively).

contralateral stance phase alterations to enable foot clearance.

The current study describes initial testing of an IMU and treadmill based gait retraining program. Whereas previously described systems have used camera-based motion tracking, use of IMUs shows increased potential to augment both clinical and home-based gait training programs [37,38]. The study provided evidence that pediatric participants can interpret and respond to visual feedback on knee flexion patterns to realize intended changes in gait patterns; however, a sub-analysis of older and younger age groups indicated the importance of understanding developmental factors in the design of feedback interventions for pediatrics. In contrast with this study, clinical use of VKFS training would occur in repeated practice across multiple sessions, promoting progression into the third phase of motor learning [32]. Moreover, this approach would entail targeting more normative gait patterns rather

Table 1
Age, Sex, MACE, and PK (mean ± SD) for subgroups with ages above and below the median age. * indicates statistically significant differences from an independent *t*-test at $p < 0.05$. Lower MACE and PK values in the older group represent a trend of improving performance with age.

Subgroup	Age (yr)	Gender	TGC + 40		TGC + 20		TGC-20		TGC-40	
			MACE (°)	PK (°)	MACE(°)	PK (°)	MACE (°)	PK (°)	MACE (°)	PK (°)
Younger	9.7 ± 1.6	3M; 3F	9.0 ± 2.6	9.3 ± 4.5	9.1 ± 3.3	10.3 ± 4.7	8.8 ± 4.5	13.2 ± 4.3	10.2 ± 3.7	14.8 ± 7.3
Older	14.2 ± 1.3	3M; 3F	6.4 ± 1.6	5.2 ± 4.2	5.3 ± 1.4	4.3 ± 1.8	7.6 ± 3.5	9.0 ± 3.1	8.8 ± 4.1	12.7 ± 5.3
p	< 0.01*		0.06	0.14	0.03*	0.02*	0.64	0.08	0.55	0.59

than the artificial targets we studied, perhaps engendering more symmetric and efficient gait patterns, and representing a fundamentally more 'natural' adaptation. However, this is likely to be highly dependent on specifics of functional limitations. In the pediatric CP population, children with persistent gait deviations following successful clinical management of spasticity and contractures may particularly benefit from feedback training. It has been shown that maladaptive kinematic changes at other joints can occur in response to training based on single-joint feedback [26]. Feedback from IMU tracking of multiple joints is technically feasible but also presents further challenges to feedback presentation, cognitive integration, and response. Future work may address these overall questions on targeted use of IMU-based gait retraining to effect clinically significant functional improvements in gait function, with the potential to ameliorate the impact of gait deviations in pediatric populations.

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Declaration of Competing Interest

All co-authors are co-inventors in a pending patent application for a feedback system that is different from but has some aspects in common with the feedback system described in this paper.

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