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Lecture

Matching walking speed of controls affects identification of gait deviations in patients with a total knee replacement

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ABSTRACT

Background: The assessment of functional recovery of patients after a total knee replacement includes the quantification of gait deviations. Comparisons to comfortable gait of healthy controls may incorrectly suggest biomechanical gait deviations, since the usually lower walking speed of patients already causes biomechanical differences. Moreover, taking peak values as parameter might not be sensitive to actual differences. Therefore, this study investigates the effect of matching walking speed and full-waveform versus discrete analyses.

Methods: Gait biomechanics of 25 knee replacement patients were compared to 22 controls in two ways: uncorrected and corrected for walking speed employing principal component analyses, to reconstruct control gait biomechanics at walking speeds matched to the patients. Ankle, knee and hip kinematics and kinetics were compared over the full gait cycle using statistical parametric mapping against using peak values.

Findings: All joint kinematics and kinetics gait data were impacted by applying walking speed correction, especially the kinetics of the knee. The lower control walking speeds used for reference generally reduced the magnitude of differences between patient and control gait, however some were enlarged. Full-waveform analysis identified greater deviating gait cycle regions beyond the peaks, but did not make peak value analyses redundant.

Interpretation: Matching walking speed of controls affects identification of gait deviations in patients with a total knee replacement, reducing deviations confounded by walking speed and revealing hidden gait deviations related to possible compensations. Full-waveform analysis should be used along peak values for a comprehensive quantification of differences in gait biomechanics.

1. Introduction

To determine whether patients with a total knee replacement (TKR) achieve full recovery of gait function after their surgery, gait biomechanics is usually compared to controls. At least fifteen studies have been published in the past nine years comparing gait biomechanics at own comfortable walking speed of TKR patients and controls (Abdel et al., 2014; Alnahdi et al., 2011; Benedetti et al., 2003; Bonnefoy-Mazure et al., 2016; Lee et al., 2015; Levinger et al., 2013; Mandeville et al., 2007; McClelland et al., 2018; Metcalfe et al., 2013; Nagura et al., 2017; Paterson et al., 2018; Urwin et al., 2014; Vahtrik et al., 2014; Worsley et al., 2013; Yoshida et al., 2012). However, these comparisons at different walking speeds may incorrectly identify TKR gait deviations, because the different walking speeds will already affect its biomechanics, such as joint angles, moments and powers (Stoquart et al., 2008).

TKR patients often walk slower compared to controls (Alnahdi et al., 2011; Benedetti et al., 2003; Lee, 2016; Mandeville et al., 2007; McClelland et al., 2018; Nagura et al., 2017; Paterson et al., 2018; Urwin et al., 2014; Vahtrik et al., 2014; Xu et al., 2010). Walking speed strongly affects the magnitude and shape of gait patterns (Stoquart et al., 2008; Winter, 1983). For instance, the knee flexion angle peak during stance

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decreases with lower speed, because less shock absorption is needed (Lelas et al., 2003) and is lower in TKR patients (Benedetti et al., 2003; Bonnefoy-Mazure et al., 2016; Mandeville et al., 2007; McClelland et al., 2017). Moreover, angular velocities decrease with lower walking speed, decreasing power peak values in all lower limb joints (Winter, 1983). Joint powers have been found to be lower in TKR patients compared to controls (Lee, 2016; Levinger et al., 2013; Vahtrik et al., 2014). Therefore, reported gait deviations in TKR compared to controls might simply be caused by the effect of a lower walking speed on gait biomechanics instead of 'real' biomechanical deviations due to pathology. To identify gait deviations in TKR patients that are unrelated to walking speed and that may affect functioning of the knee affecting daily life, gait pattern comparisons need to be based on matched walking speeds.

The most straightforward method to use matched walking speeds, is to record gait patterns at preset imposed speeds. However, patients may not have the motor control to adjust gait efficiently to imposed walking speeds, or may not even be capable to do so. Therefore, previous studies compared groups at their own CWS and used speed as a covariate within an analyses of covariance (Alnahdi et al., 2011; McClelland et al., 2018; Paterson et al., 2018). However, this statistical method necessitates the use of discrete parameters, for which a priori decisions must be made on which discrete parameters will be analyzed. So other potential relevant information in the gait cycle is discarded that way. Moreover, the use of the conventional linear regression approach assumes a linear relation between the covariate (speed) and the dependent variable (discrete parameters) for both controls and TKR patients, which is often incorrect (Lelas et al., 2003). Gait patterns instead can be described through waveforms in which different features, such as amplitude and timing, can vary between conditions (group or task). Several aspects of waveforms, such as timing, are not easily captured with discrete parameters only. To be able to account for walking speed and analyze gait pattern changes over the full waveform, full gait cycle data that can be corrected for walking speed is needed.

To facilitate a matched walking speed gait comparison over the full gait cycle, control gait data could be recorded at all walking speeds of the patients to which they are compared. However, this would necessitate the collection of a new norm dataset for each comparison study. Therefore, based on Glardon et al. (2004) (Glardon et al., 2004) we have developed a method to reconstruct full biomechanical gait waveforms of controls matched to every possible speed within a patient population, informed by a norm dataset of controls at multiple fixed walking speeds (Meinders et al., 2019). This method allows patients to walk at their comfortable walking speed, while the walking speed of controls is synthesized. This new walking speed correction method allows analyses of full waveforms. Statistical parametric mapping (SPM) is a suitable method to conclude any differences over the full waveform (Pataky, 2010). This method identifies the regions over the gait cycle that differ significantly between groups.

The aim of this study is to 1) describe the effect of not-matching vs. matching walking speeds on the identified gait deviations, 2) investigate the additional value of full-waveform analyses in identification of TKR gait deviations in comparison to conventional peak value analyses.

We hypothesize that correcting control gait data for walking speed will reduce the amount of gait deviations in TKR patient gait compared to uncorrected control gait, since TKR patients tend to walk slower than controls. Specifically, we expect to find the strongest effect of walking speed correction on joint power outcomes (Winter, 1983), specifically on the knee joint (Chen et al., 1997), while the smallest effect is expected on kinematic outcomes (Lelas et al., 2003; Winter, 1983). Further, we expect that full-waveform analyses offer a more complete assessment of TKR gait deviations over discrete measures, since they take all biomechanical information into account.

2. Methods

2.1. Participants

Lower limb gait waveforms were compared between 25 patients one to five years after receiving their primary TKR and 22 controls within the same age range. All participants were between 50 and 75 years old, could walk without aids and had no comorbidities that could affect the gait pattern, such as diagnosed knee osteoarthritis or a prosthesis in any lower limb joint (other than the knee replacement for patients). Ethics approval was granted from the local Human Research Ethics Committees (NL51829.029.14), and written informed consent was provided according to the Committees' guidelines.

2.2. Data collection

Gait parameters were collected on the GRAIL (Gait Real-time Analysis Interactive Lab, Motek ForceLink BV, Netherlands) at the rehabilitation department of the Amsterdam University Medical Center, location VUmc. Participants walked on an instrumented treadmill. During the walking trials, 3D motion was captured via InfraRed optical motion capture with wireless, light-reflecting markers (Vicon version 2.5, Oxford, UK). For this study, 26 markers were placed on the subject for reconstruction of the position and orientation of the lower limbs, pelvis and trunk in space, according to the CAST model (Cappozzo et al., 1995), recorded at a 100 Hz sample frequency. Additionally, ground reaction forces were measured using two 6D force plates (KISTLER type 9281, Kistler Instrumente AG, Winterthur, Switzerland), recorded at a 1000 Hz sample frequency.

After a minimum of 5 min of habituation to treadmill walking, gait data were recorded at several walking speeds. TKR patients walked at their comfortable walking speed (CWS). The asymptomatic controls walked at five speeds: CWS and four speeds that were equally distributed over the range of non-dimensional speeds at which the TKR patients walked, i.e. 0.2–0.5. Additional habituation time to non-CWS walking speeds were provided, with a minimum of three minutes. Non-dimensional walking speeds of the patients were calculated by normalizing the absolute speed in m/s for leg length with eq. 1 (Hof, 1996). Then the absolute walking speeds for the controls were calculated using their own leg length.

$$v_{nor} = \frac{v}{\sqrt{(g \cdot L)}} \tag{1}$$

with v fixed walking speed of subject, v_{nor} normalized walking speed (v_{nor} = [0.2, 0.3, 0.4 and 0.5] for controls), g gravitational acceleration, and L leg length.

2.3. Data processing

Marker and force plate data were filtered using a two-way second order low pass Butterworth filter with cut-off frequency of 6 Hz. Inverse kinematics and kinetics were performed using custom-made software BodyMech (www.bodymech.nl). All joint moments were expressed externally and in the distal segment coordinate frame and normalized to body weight. All data were time-normalized, such that one full gait cycle was represented by 100-time samples.

Fifteen gait variables were analyzed in this study: sagittal and frontal kinematic and kinetic gait variables (joint angles, moments and overall powers) for the three major lower limb joints (hip, knee, ankle). The gait waveforms on the replaced side of the patients and both sides for controls were analyzed.

2.4. PCA correction for walking speed

To correct the gait waveforms of the control data for walking speed, a

PCA correction was applied, PCA is known to be reliable to in gait analysis data (Robbins et al., 2013). Gait waveforms of controls were reconstructed to match the walking speed of each TKR patient (25 patients, thus 25 reconstructions) using PCA. For each gait waveform, principal components (PCs) were extracted from the full dataset of all controls at five walking speeds, using twelve strides. These twelve strides were the closest to the median strides per speed based on the knee flexion moment. The dataset was a matrix of 1320 (22 controls $\times$ 5 speeds $\times$ 12 strides) x 100 (time samples in a gait cycle), which formed the input for the PCA. The PCs explaining in total 90% of the total variance from the PCA were included for use in the speed correction. Using a stepwise linear regression model, the relation between the PC-scores and walking speed was determined for each PC. Based on this relation, new PC-scores at each of the 25 patient walking speeds were estimated. These new PC-scores and the corresponding PCs were used to reconstruct control gait at each patient speed. This resulted in a new control dataset for each gait waveform consisting of 25 waveforms that each match the speed of the 25 patients (Meinders et al., 2019).

After the speed correction gait waveforms from three groups were compared: 1) controls at their own CWS (CON), 2) controls corrected towards matched TKR speeds (matched CON), and 3) TKR patients at their CWS (TKR).

2.5. Discrete parameters

Peak values from the gait waveforms were calculated for all three groups (CON, matched CON, TKR). The peak values analyzed for angles and moments were the most common discrete parameters (Fig. 3 & Table Appendix A) used in comparison studies between TKR patients and controls (Alnahdi et al., 2011; Benedetti et al., 2003; Bonnefoy-Mazure et al., 2016; Lee, 2016; Levinger et al., 2013; McClelland et al., 2018; Metcalfe et al., 2013, 2017; Nagura et al., 2017; Paterson et al., 2018; Smith et al., 2006, 2004; Urwin et al., 2014; Vahtrik et al., 2014; Worsley et al., 2013; Xu et al., 2010; Yoshida et al., 2012). Additionally, peak power parameters were based on Winter (Winter, 1983). Two to four peak values were calculated per gait waveform for each group.

Outcomes were described using the following gait phases: early stance (1–20% of the gait cycle), midstance (20–40%), late stance (40%-toe-off (±66%)), and early swing (toe-off −75%).

2.6. Statistical analysis

First, the gait waveforms were selected that showed a clear group effect of speed using the control data at the four fixed speeds. If the variation of the gait pattern between speeds was too large, the response to a change in speed cannot be predicted. Therefore, a speed distinction ratio was calculated. Speed distinction ratio: the average standard deviation between subjects over time averaged over all speeds (group variation) divided by the difference in average amplitude of the waveforms over time between subsequent fixed speeds averaged over all speeds (speed effect). If the speed distinction ratio was larger than 10 (the group variation was 10 times larger than the average speed effect), the waveform was excluded for further analyses.

Second, to investigate the effect of the PCA correction of walking speed, gait waveforms were compared between TKR patients and controls. The value of the peak parameters were compared using Student's *t*-tests: 1) uncorrected model (TKR vs. CON) and 2) the PCA corrected model (TKR vs. matched CON). These group comparisons were also performed using t-tests within statistical parametric mapping (SPM) (Pataky, 2010) for full-waveform analyses, to identify regions over the gait cycle that differ significantly between these groups. Both statistical methods (peak and full-waveform analyses) were compared using the significant findings and their timing in the gait cycle at which they occurred.

Third, differences between the comparison of TKR to CON versus matched CON were calculated, to quantify the effect of the speed correction on these deviations (Fig. 1). For each gait waveform SPM was used to show which regions in the gait cycle were significantly different between groups (Fig. 1A). For these regions, additional knowledge of the relative magnitude of the deviation of TKR gait is of interest. Therefore, only the gait regions identified as significantly different between TKR and either CON or matched CON, were used for quantification of the speed correction effect. For kinematic data, the mean angle for each group was determined over these significant deviating gait cycle regions identified through SPM. Then, the absolute difference between the mean angles of TKR and the control groups were calculated. For the kinetic data, the absolute area under the gait waveforms of the three groups was calculated for the significant TKR deviating regions (Fig. 1B-D). The areas were calculated separately for moments in each direction (adduction vs. abduction and flexion vs. extension) and for power generation and absorption. These areas were summed per waveform and the TKR values were expressed as a percentage of the absolute area under the curve of the (matched) control waveform for the same regions (Fig. 1C&D). These percentages show per waveform how much TKR gait is deviating from controls for the corrected and uncorrected control data. Finally, the effect of the speed correction was quantified by calculating the difference between the corrected and uncorrected magnitude of the TKR deviations. If the effect of the speed correction is negative, the magnitude of the TKR deviation is reduced by application of the speed correction. The larger the effect, the more sensitive TKR deviations in this waveform are for the speed correction.

All analyses were carried out in MATLAB R2018b (MathWorks, Natick, MA). Significance was set at alpha 0.05.

3. Results

3.1. Subject characteristics

The TKR group had significant higher BMI than the control group (mean difference of 3.2 kg/m2, $p < 0.01$) and significant lower CWS (0.22 m/s, $p = 0.001$)(Table 1).

3.2. Selection gait waveforms

As shown by the speed distinction ratio, joint adduction angles and ankle adduction moment were not related to changes in walking speed (Fig. 2, Table 2). Little or no differences were observed in these gait waveforms over speed, with high variation between subjects. Therefore, speed correction seemed inappropriate for these gait variables and were consequently not considered for further analyses.

3.3. Effect of the speed correction on TKR gait deviations

Significant group differences between TKR and controls remained for all knee variables after speed correction, except the knee power absorption in late stance (Fig. 3). Furthermore, the hip and ankle flexion moments and powers showed several differences in early and late stance which were no longer significantly different after speed correction. Most group differences were found during late stance in all gait waveforms, with newly identified differences in late stance and early swing in the hip flexion moment and hip and knee powers after speed correction.

To quantify the effect of the speed correction, the magnitude of TKR deviation in deviating gait cycle regions - as identified by full-waveform analyses (SPM) - were compared before and after applying the correction (Fig. 4).

SPM showed no significant differences in the hip flexion angle between TKR patients and control data (Figs. 3 and 4) and therefore no speed correction effect on this gait variable was shown (Fig. 4, Table 3). In all other gait waveforms an effect of the speed correction was present.

The speed correction generally reduced the difference between TKR and control gait patterns, as seen in negative numbers of the speed correction effect (Table 3). TKR knee adduction moment deviation

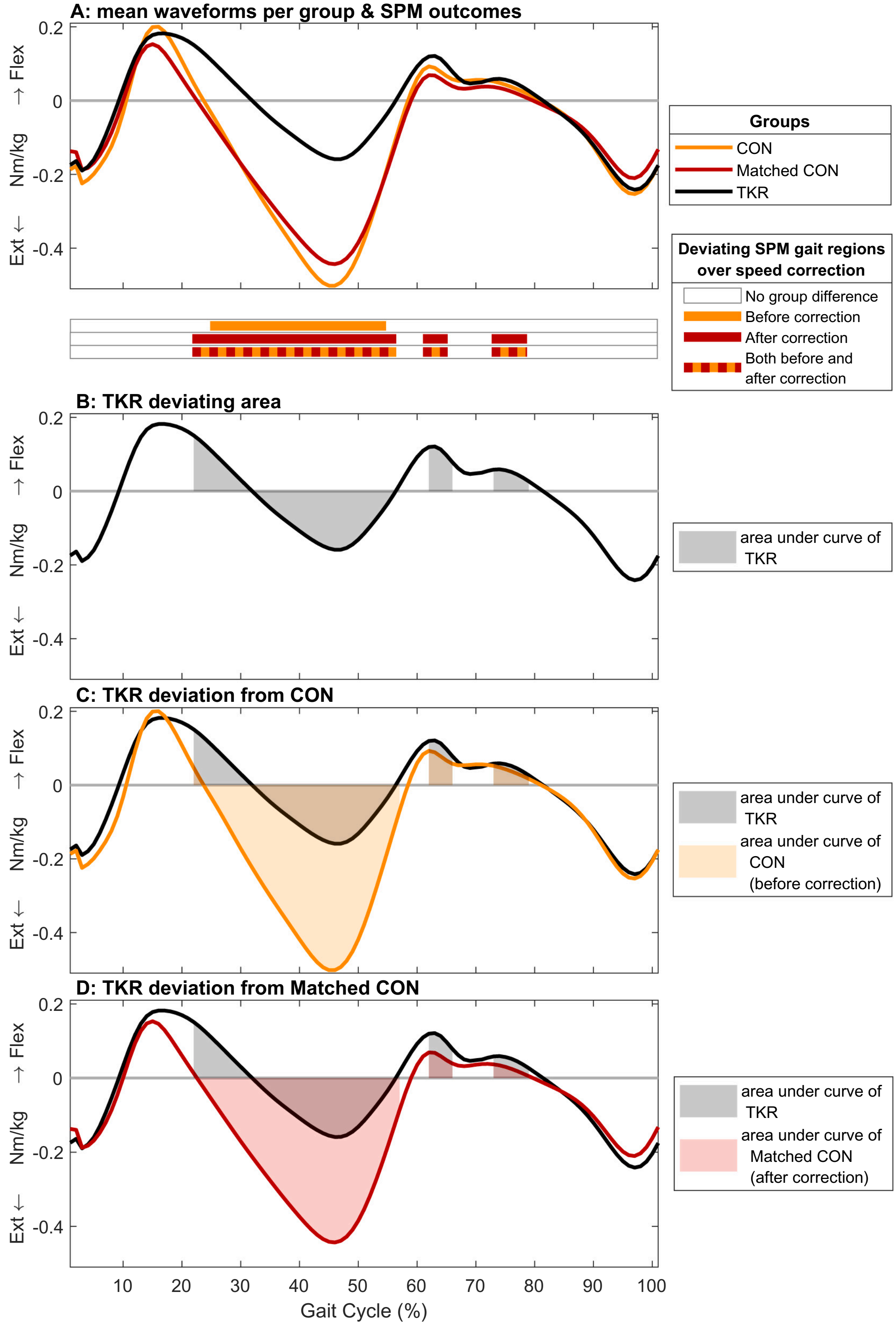

Fig. 1. Method: quantification speed correction effect The knee flexion moment is visualized as an example. A: Using SPM deviating TKR regions are identified: the regions of the TKR gait cycle that differ significantly from before (CON, orange) and after (Matched CON, red) the speed correction, then combined (dashed orange/red) as shown in the bars below the graph. B: The sum of the areas under the deviating TKR curve is calculated (grey area). C,D: The area under the curve of CON (C) and Matched CON (D) are calculated for the TKR deviating gait regions. The TKR areas are expressed as a percentage of the area under the curve of the control gait waveforms. This calculation is performed separately for the areas above zero (above the grey line, flexion moment) and below zero (extension moment). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 1
Demographics Mean ± SD.

	N	Gender	Age (years)	BMI (kg/m^2)	Comfortable Walking Speed (m/s)	Non-dimensional walking speed
TKR	25	16 male	63.8 ± 5.1	29.0 ± 2.9	0.98 ± 0.29	0.32 ± 0.09
CON	22	13 male	66.5 ± 5.2	25.8 ± 3.2	1.20 ± 0.16	0.40 ± 0.05

decreased the most with 38% after speed correction from 203% to 164% compared to controls. Furthermore, large effects of the speed correction were shown for the hip joint power generation and the knee and ankle power absorption (−26, −21 and − 23% respectively). TKR deviations in the hip extension and knee flexion moments increased strongly after speed correction, with 37% and 120% respectively.

3.4. Different identified TKR deviations using peak vs. full-waveform analyses

Several differences in the knee between controls and TKR were identified by both the analyses of peaks and gait cycle regions using SPM. Knee extension angle peak differences were found at 43 and 44% of the gait cycle (4.6–6.4 degrees difference) and peak ankle dorsal flexion angle differences were found at 52 and 53% of the gait cycle (3.4 degrees difference), while SPM showed differences from 31 to 48% and 53–67% of the gait cycle, respectively. Knee flexion moment peak differences were found at 46% of the gait cycle (0.27–0.33 Nm/kg difference), while SPM showed group differences from 22 to 57% and 62–66% of the gait cycle (Fig. 3, Table Appendix A). Furthermore, the peak knee adduction moment in midstance and late stance were significantly different between groups around 40 and 50% of the gait cycle (0.12–0.15 Nm/kg), while SPM showed group differences from 25 to 66% of the gait cycle.

Significant peak differences in the hip and ankle waveforms were found, while SPM did not find any significant differences at the same time points. TKR patients showed lower early stance peak hip flexion moment (0.15 Nm/kg mean difference, occurring at 10.9% for TKR and 11.2% for controls), lower plantar ankle flexion moment (0.028 Nm/kg difference, occurring at 5.8% for TKR and 6.0% for controls) and lower ankle power absorption (0.36 W/kg difference, occurring at 49.0% for TKR and 48.8% for CON), before but not after speed correction was applied. These peak differences were not identified by SPM. Remaining higher hip adduction moment peaks for TKR patients after speed correction (0.13 Nm/kg difference, occurring at 48.9% for TKR, 47.9% for CON and 47.1% for Matched CON) and greater TKR hip power peaks (late stance absorption difference 0.21 W/kg, at 55% for TKR, 54% for matched CON and early swing generation difference 0.16 W/kg at 73% for TKR, 71% for Matched CON) were identified with peak parameters, but not with SPM.

Significant deviating gait cycle regions were found by SPM, while peak parameter analyses did not find significant differences for these regions. For example, in the knee power the gait waveform was significantly different between TKR and matched CON at 25–41% and 44–56%, while no peaks were calculated in these regions. Moreover, for most gait variables SPM showed in the region around toe-off, between 60 and 70% of the gait cycle, significant differences between the groups, with no peaks shown in this region.

4. Discussion

Matching gait speed of controls to patients generally reduced the identified TKR gait deviations, while some deviations became more apparent. These results show the importance to consider the appropriate walking speed for gait biomechanical waveform comparisons between controls and TKR patients. Moreover, full-waveform analyses identified more and sometimes different TKR gait deviations compared to peak value analyses. These new insights may impact conclusions drawn regarding TKR outcome assessment and indications for training programs.

4.1. Effect of the speed correction

All joint variables were impacted by the speed correction, except for the hip angle, adduction angles and ankle adduction moment. As expected, the speed correction generally reduced gait deviations in TKR patients. The speed correction affected the magnitude of the TKR gait deviations most in the knee moment waveforms, followed by the hip extension moment and powers.

In previous studies little variation in joint angles with walking speed was shown (Lelas et al., 2003; Winter, 1983). Our study did show clear effects of walking speed on flexion angles for all joints (Fig. 1), which were measured over a larger range of speeds than previous studies. However, the speed correction did have little impact on the change in TKR deviation considering the joint angle range. The mean TKR joint angle deviation was less than 5% of the ankle angle range during gait, which was almost negligible in the knee (0.5%) (Table 3). This is in line with our hypothesis.

In agreement with previous literature, joint moment variables were clearly influenced by walking speed (Lelas et al., 2003). The hip extension and knee flexion moments showed the strongest increase in TKR deviation after the speed correction (37% and 120% respectively).

As expected, the joint power waveforms of control gait were influenced by speed (Chen et al., 1997). Consequently, all joint power waveforms were clearly impacted by the speed correction. However, in contrast to our hypothesis, the knee power did not show the strongest effect to the speed correction but was equally affected as the hip and ankle powers. This difference was probably caused by the relative expression of only the deviating gait cycle area, while Chen et al. expressed the absolute area under the full waveform of the knee joint power (Chen et al., 1997). Further, a lower early stance peak hip power generation was found for TKR patients before speed correction, in agreement with Lee et al.[17], while this finding was no longer significant after speed correction. Additionally, hip power absorption started significantly earlier for TKR patients (39–45% gait cycle, Fig. 4), while controls still absorbed hip power in this region of the gait cycle and the following peak power absorption was greater in TKR patients.

The emergence of the greater hip extension moment peak and late stance and early stance power TKR deviations after the speed correction (Fig. 3) are new findings for this population. These new findings show that the correction can also enlarge gait deviations in a comparison between gait patterns, shedding light on possible compensation strategies. While other TKR gait deviations identified in previous research showed to be speed dependent, because they were removed when walking speeds were matched.

In summary, the greatest impact of the speed correction was seen in the kinetic gait data.

There is no single joint or variable that was most impacted by the speed correction. Moreover, identified gait deviations did change applying the speed correction. Therefore, it is advised to always correct for walking speed when comparing gait patterns between groups, especially when kinetic gait data are compared to controls.

Previous clinical research on gait deviations in TKR patients has paid little attention to several variables that were analyzed in this study, such as knee extension angle and knee and hip extension moment and power. Causes for these deviations may lie in patients' motor control, as increased knee flexion has been related to a cautious gait pattern (Henderson et al., 2019), which could be attributed to differences in proprioception, muscle activation and co-contraction patterns in TKR patients (Smith et al., 2006). However, pain symptoms also potentially

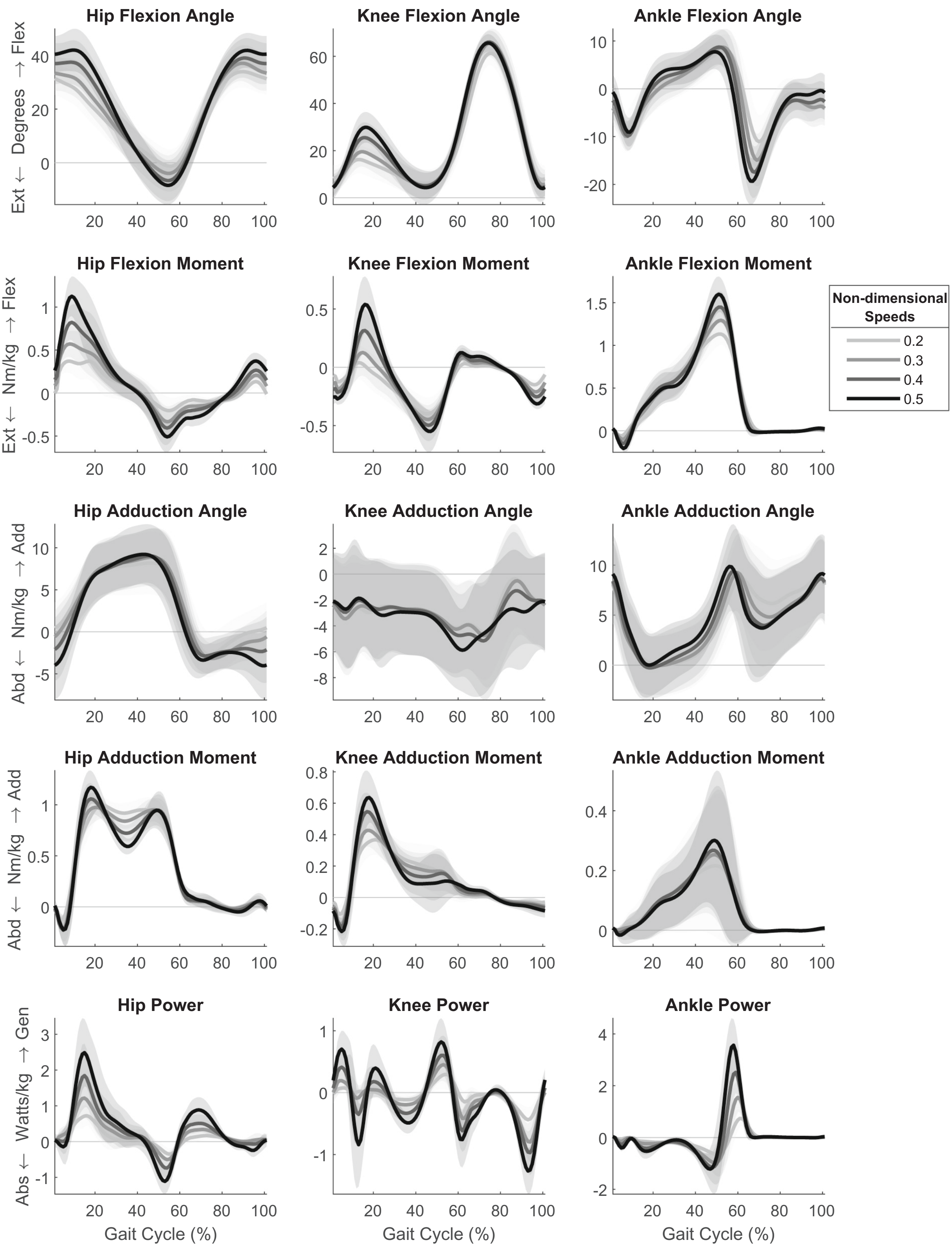

Fig. 2. Speed effect Darker lines represent mean gait waveforms at higher walking speeds. The light grey error band shows one standard deviation from the mean. The gait waveforms show a relation to speed when the mean lines at different speeds clearly deviate from each other. A lower standard deviation gives more certainty on the presence and type of the relation to speed. The adduction angles and the ankle adduction moment waveforms do not clearly deviate between gait speeds and show a large standard deviation. Therefore, these gait waveforms are excluded from further analyses.

Table 2
Speed distinction ratio.

		Speed distinction ratio
Hip	Flexion Angle	7.0
	Adduction Angle †	10.3
	Flexion Moment	4.2
	Adduction Moment	5.2
	Power	2.6
Knee	Flexion Angle	3.1
	Adduction Angle †	22.4
	Flexion Moment	7.0
	Adduction Moment	9.8
	Power	4.5
Ankle	Flexion Angle	4.5
	Adduction Angle †	33.6
	Flexion Moment	8.6
	Adduction Moment †	12.2
	Power	8.2

† Excluded variables, because of ratio > 10.

cause increased knee flexion (Smith et al., 2004). Subsequently, the increased hip extension moment and power may show compensations for this knee dysfunction (Mcgibbon and Krebs, 2002). Future research should investigate whether these newly identified gait deviations are relevant targets for rehabilitation programs.

4.2. Peak and full-waveform comparisons are complementary

While peak parameters are the conventional way to assess gait biomechanics (Sosdian et al., 2014), it ignores a lot of information in the gait pattern. To account for this, we additionally compared gait waveforms using SPM for full-waveform analyses (Pataky, 2010).

As expected, the application of full-waveform analyses identified greater regions of deviating gait than with peak analyses, clearly shown in the knee angle and moments (Fig. 3). Moreover, full-waveform analyses do not need previous knowledge of relevant peak parameters, e.g. the absence of peak parameters calculated in the deviating region of the knee power (25–56% gait cycle). Further, SPM showed differences in regions around toe-off in multiple waveforms, indicating a phase shift (i.e. longer stance phase), not captured by peak parameters.

On the other hand, several differences in peak parameters of the gait pattern were identified but missed by SPM, such as peak hip power in late stance and early swing. This is due to insensitivity of peak parameters to the exact timing of the peaks, in contrast to SPM, that is based on a common independent time-base.

Full-waveform analysis facilitates to look beyond known gait deviations in TKR patients. Especially deviations other than the amplitude of peak joint angle, moment, or power. This is related to the concept of the recently upcoming interest in summation gait parameters such as the impulse, but SPM makes the outcome more concrete. Therefore, full-waveform analysis allows for identification of novel targets for rehabilitation programs.

In conclusion, full-waveform analyses provided more information on the TKR gait deviations than using peak parameters. Further, full-waveform analyses are less affected by prior choices, such as the selection of which peaks to include in the analyses. This emphasizes the benefit of the full-waveform PCA speed correction. Thus, full-waveform analyses are the preferred method for comparison of gait patterns. However, due to the additional value of the peak parameters, the methods should be combined for a complete overview of the gait deviations.

4.3. Limitations

This is the first study in which the new speed correction method was applied to a large gait biomechanical dataset in patients and controls. This correction method has only been validated for the knee flexion and adduction moments, not for kinematics and powers (Meinders et al., under review). However, it is very likely that this validation can be translated to the other outcome measures analyzed in this study.

Two aspects of the current study resulted in an elevated chance of type I errors; the method and the number of tests. The PCA reconstruction method removes noise from the control dataset, because only the principal components explaining in total 90% of the variance are included, leading to a smaller value-to-noise ratio in the matched CON dataset. However, the outcomes of the regression analyses (Table Appendix A) showed significant differences due to an increase in effect size and not a reduction in standard error. Therefore, the differences in outcomes due to application of the speed correction can be attributed to a difference in walking speed between the control groups. Secondly, a lot of tests were performed. The aim of this study is of explorative nature. Its results should improve the methodology of future studies and give insight into which outcomes may be of importance for future studies on gait of the TKR population. Therefore, possible false positive findings due to type I errors should be eliminated by future research.

An important practical consideration of the speed correction method is the necessity to have control gait data recorded at a large range of speeds. Therefore, most available norm datasets will need to be extended to a wider range of speeds. However, this dataset will allow matching walking speeds to any patient. Therefore, the correction method provides more feasible future comparison studies using matched speeds.

Finally, we have only reported the data of the operated limb of the TKR patients. It is possible that patients show compensations in the non-operated side, which can well be investigated with the same method as applied in this study. However, in line with previous research (Yoshida et al., 2012), we found minimal interlimb differences at this point in the rehabilitation process.

5. Conclusion

Using biomechanical reference data at matched speeds affects gait deviations when comparing TKR patients with controls, especially when studying kinetic gait data. Most TKR deviations were reduced in size, showing their dependency on walking speed. However, some deviations were revealed, demonstrating possible compensation strategies. Moreover, a correction method that includes full waveforms, shows clear additional value to merely speed corrected discrete parameters.

Therefore, it is advised to perform future studies comparing gait patterns of different groups or in different studies on equal walking speeds. Correcting walking speed with the speed correction method facilitates more feasible comparison studies, allowing for more efficient walking speed matching. Future studies may focus on TKR deviations that remained significant after the speed correction. Causes for these compensations and their relevance for rehabilitation therapies need to be investigated.

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Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

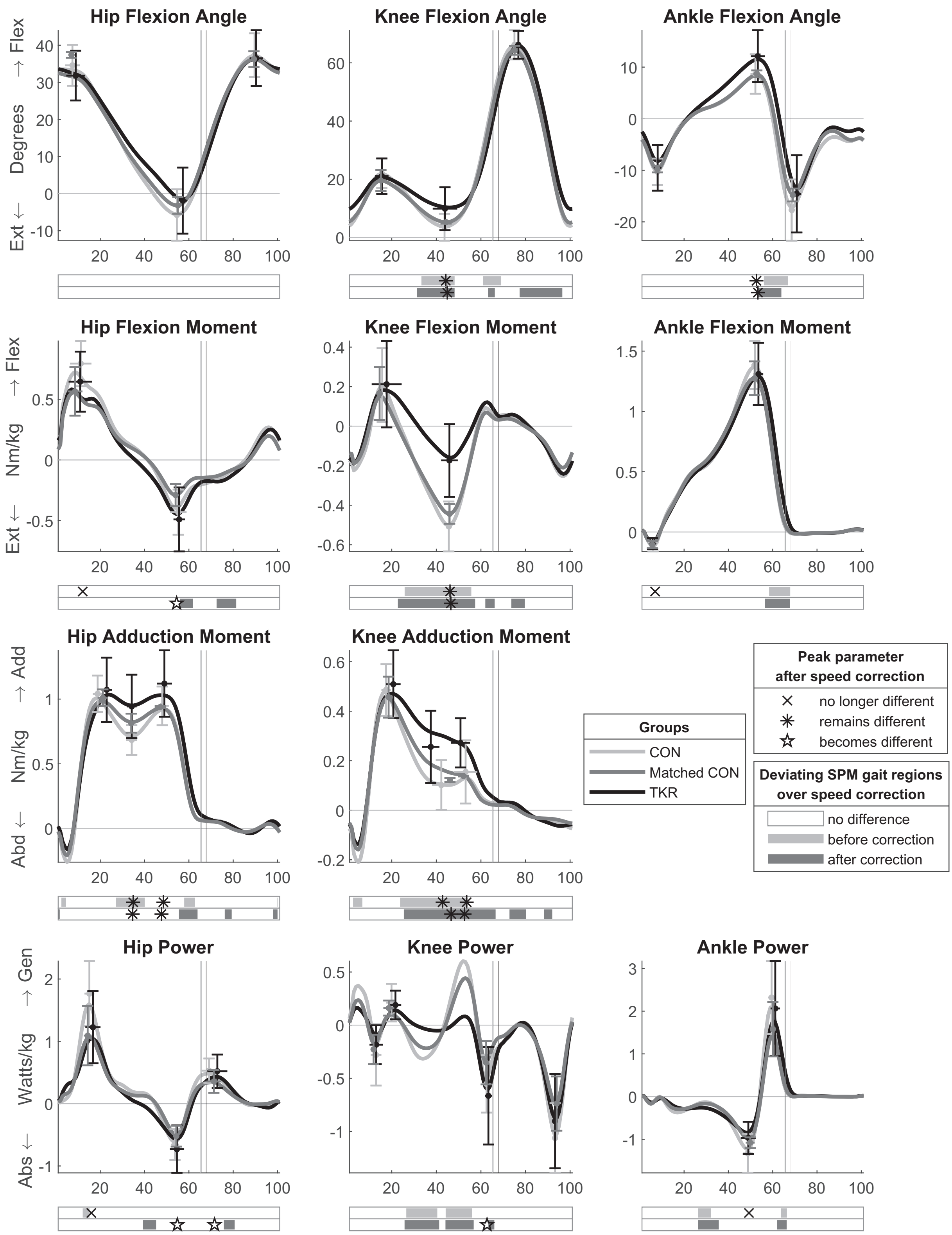

Fig. 3. Gait waveforms and discrete parameters of the three datasets: TKR patients (black), recorded controls (CON, light grey), speed matched control data (Matched CON, dark grey). Peak values are shown with standard deviations as whiskers over time and amplitude. Timing of toe-off is shown in vertical lines. Below the graphs deviating gait cycle regions identified with SPM are shown in the horizontal bars; light grey lines show the differences between TKR and CON and dark grey lines between TKR and Matched CON.
Markers show significant differences only in the comparison uncorrected for speed (×), only in the speed corrected comparison (☆), or both (✳) at significant level α = 0.05.

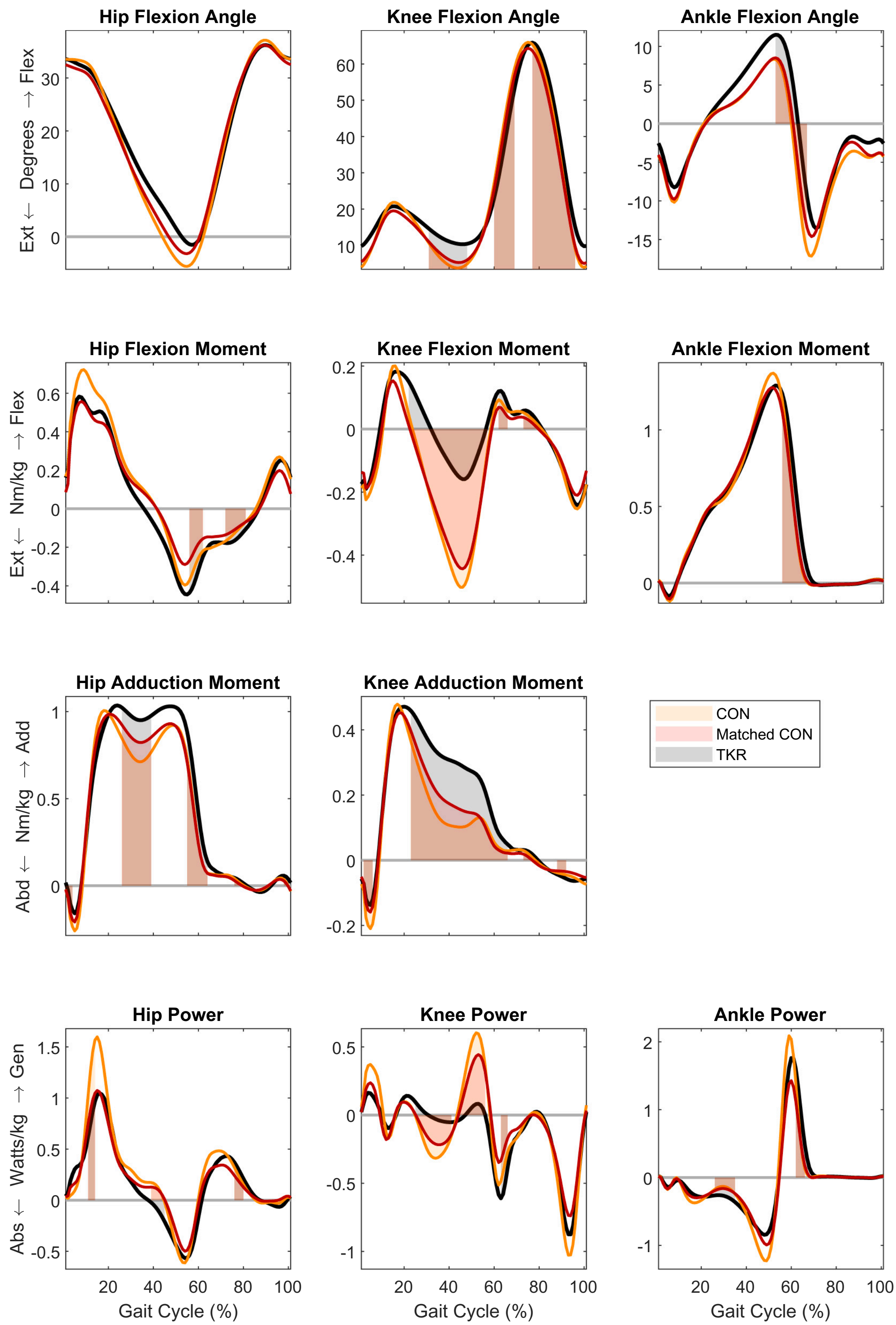

Fig. 4. Quantification speed correction effect visualized using area under the curve for the phases of the TKR gait cycle identified as significant different from controls. Total area of deviating TKR gait (grey) is expressed as a percentage of the area under the curve of CON (orange) and Matched CON (red) for the same gait cycle regions. Large filled TKR areas (grey areas), without overlap of control gait areas (orange or red areas) indicate large deviation of the TKR gait pattern, as well as vice versa the control areas that do not overlap the TKR areas. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 3
Quantification of the speed correction effect on TKR deviations expressed over the significant SPM gait cycle region differences between TKR and controls, as mean angle difference for the kinematic data (A) and as a percentage of area under the curve of the control gait waveforms for kinetic data (B). A positive value of the effect of the speed correction expresses an increase in TKR deviation after the correction, a negative effect expresses a decrease in deviation.

A			Angle range of deviating TKR gait waveform	TKR deviating gait angles (absolute mean degrees difference to control; TKR - CON)		Effect speed correction (degrees)
				Uncorrected	Corrected	Corrected - Uncorrected
Hip	Angle	Flex&Ext	–	–	–	–
Knee	Angle	Flex&Ext	55.6	5.0	4.7	−0.3
Ankle	Angle	Flex&Ext	34.2	6.1	4.3	−1.7

B			Area under curve of the deviating TKR gait waveform	Control gait cycle region of deviating TKR gait (% of area under the deviation TKR waveform)		Effect speed correction (%)
				Uncorrected	Corrected	Difference between Corrected and Uncorrected
Hip	Moment	Flexion	–	–	–	–
		Extension	3.7	123.5	160.0	36.5
	Moment	Adduction	18.8	137.2	126.3	−11.0
		Abduction	0.2	36.7	43.2	−6.5
	Power	Generation	4.3	63.9	89.4	−25.6
		Absorption	0.6	Inf	Inf	0.0
Knee	Moment	Flexion	1.6	244.6	364.7	120.1
		Extension	2.3	23.4	25.1	−1.7
	Moment	Adduction	12.0	202.5	164.3	−38.2
		Abduction	0.8	75.0	95.7	−20.7
	Power	Generation	0.7	14.1	20.1	−6.0
		Absorption	2.4	46.4	67.3	−20.9
Ankle	Moment	Flexion	7.5	136.2	139.2	3.1
		Extension	–	–	–	–
	Power	Generation	4.2	158.9	172.1	13.2
		Absorption	2.6	180.7	158.1	−22.6

Appendix

Appendix Table A
Peak parameters are either a positive or negative peak in the early stance (es), midstance (ms), late stance (ls) and swing (sw) phase. Peak parameters of the powers are indicated with standard names from literature (Winter, 1983). Mean and standard deviation (SD) values of the discrete parameters for the gait data of the controls (CON), corrected data of the controls (Matched CON) and for the TKR patients (TKR) are shown. T-test outcomes are shown for TKR patients to controls uncorrected and speed corrected; mean difference (Mean diff), standard error (SE) and p-value. Significant group differences are indicated in bold numbers.

			Markers	CON		Matched CON		TKR		Uncorrected (CON vs TKR)			Speed Corrected (Matched CON vs TKR)		
				Mean	SD	Mean	SD	Mean	SD	Mean diff	SE	p-value	Mean diff	SE	p-value
Hip	Flexion Angle	peak es		34.60	5.53	37.42	0.78	31.82	6.71	−2.777	2.313	0.241	−5.596	4.909	0.277
		peak ls		−5.75	6.94	−3.19	2.25	−1.88	8.90	3.871	2.351	0.107	1.311	1.835	0.479
		peak sw		37.33	5.90	36.26	2.14	36.51	7.52	−0.822	1.991	0.682	0.252	1.564	0.873
	Flexion Moment	peak es	×	0.79	0.19	0.57	0.20	0.65	0.25	−0.149	0.065	**0.026**	0.080	0.064	0.214
		peak ls	☆	−0.43	0.19	−0.29	0.09	−0.49	0.26	−0.060	0.068	0.381	−0.201	0.056	**0.001**
	Adduction Moment	peak es		1.04	0.14	1.01	0.06	1.07	0.25	0.031	0.063	0.631	0.062	0.053	0.248
		peak ms	✳	0.68	0.11	0.81	0.07	0.94	0.25	0.258	0.061	**0.000**	0.129	0.053	**0.018**
		peak ls	✳	0.95	0.15	0.94	0.01	1.12	0.26	0.172	0.064	**0.011**	0.181	0.053	**0.002**
	Power	peak H1	×	1.77	0.52	1.09	0.48	1.23	0.58	−0.540	0.161	**0.002**	0.135	0.150	0.373
		peak H2	☆	−0.66	0.25	−0.52	0.17	−0.73	0.38	−0.073	0.095	0.448	−0.209	0.083	**0.016**
		peak H3	☆	0.53	0.19	0.36	0.18	0.52	0.27	−0.010	0.069	0.891	0.164	0.065	**0.015**
Knee	Flexion Angle	peak es		22.03	5.10	19.55	3.60	21.08	6.09	−0.953	1.651	0.567	1.527	1.414	0.286
		peak ls	✳	3.49	4.62	5.24	0.51	9.89	7.38	6.395	1.826	**0.001**	4.641	1.480	**0.003**
		peak sw		66.20	5.00	64.31	1.53	66.21	4.79	0.012	1.429	0.994	1.901	1.005	0.065
	Flexion Moment	peak es		0.21	0.19	0.16	0.13	0.21	0.22	0.004	0.060	0.941	0.047	0.051	0.359
		peak ls	✳	−0.51	0.13	−0.44	0.05	−0.17	0.18	0.334	0.047	**0.000**	0.270	0.038	**0.000**
	Adduction Moment	peak es		0.49	0.10	0.46	0.08	0.51	0.14	0.023	0.036	0.520	0.052	0.032	0.110
		peak ms	✳	0.10	0.10	0.12	0.01	0.26	0.15	0.154	0.047	**0.003**	0.134	0.049	**0.014**
		peak ls	✳	0.15	0.13	0.14	0.00	0.27	0.10	0.117	0.042	**0.008**	0.134	0.030	**0.000**
	Power	peak K1		−0.28	0.29	−0.23	0.14	−0.18	0.18	0.096	0.072	0.191	0.044	0.046	0.347
		peak K2		0.21	0.18	0.16	0.07	0.19	0.13	−0.022	0.048	0.642	0.029	0.031	0.354

(continued on next page)

Appendix Table A (*continued*)

			Markers	CON		Matched CON		TKR		Uncorrected (CON vs TKR)			Speed Corrected (Matched CON vs TKR)		
				Mean	SD	Mean	SD	Mean	SD	Mean diff	SE	p-value	Mean diff	SE	p-value
		peak K3	☆	−0.55	0.27	−0.35	0.20	−0.67	0.46	−0.112	0.112	0.321	−0.311	0.100	**0.003**
		peak K4		−1.06	0.33	−0.74	0.26	−0.90	0.44	0.158	0.115	0.176	−0.166	0.102	0.113
Ankle	Flexion Angle	peak es		−10.30	2.56	−9.77	0.64	−9.53	4.43	0.770	1.080	0.480	0.240	0.895	0.790
		peak ls	✳	8.67	3.83	8.70	0.67	12.12	5.02	3.448	1.317	**0.012**	3.419	1.013	**0.001**
		peak sw		−17.71	5.94	−14.78	1.25	−14.56	7.50	3.150	1.992	0.121	0.220	1.520	0.885
	Flexion Moment	peak es	×	−0.12	0.04	−0.11	0.03	−0.10	0.05	0.028	0.012	**0.025**	0.009	0.011	0.392
		peak ls		1.38	0.20	1.27	0.14	1.31	0.26	−0.074	0.068	0.279	0.035	0.059	0.550
	Power	peak A1	×	−1.33	0.46	−1.08	0.13	−0.97	0.38	0.362	0.122	**0.005**	0.112	0.080	0.166
		peak A2		2.32	0.85	1.58	0.64	2.06	1.11	−0.257	0.292	0.382	0.483	0.256	0.065

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