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Locomotion pattern and foot pressure adjustments during gentle turns in healthy subjects

Kathrin E. Peyer[1], Charlotte Brassey[2], Kayleigh Rose[1], and William I. Sellers[1]

[1] School of Earth and Environmental Sciences, The University of Manchester

[2] School of Science and the Environment, Manchester Metropolitan University

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Corresponding author:

Kathrin E. Peyer, kathrin.peyer@manchester.ac.uk

Abstract

People suffering from locomotor impairment find turning manoeuvres more challenging than straight-ahead walking. Turning manoeuvres are estimated to comprise a substantial proportion of steps taken daily, yet research has predominantly focused on straight-line walking, meaning that the basic kinetic, kinematic and foot pressure adaptations required for turning are not as well understood. We investigated how healthy subjects adapt their locomotion patterns to accommodate walking along a gently curved trajectory (radius 2.75 m). Twenty healthy adult participants performed walking tasks at self-selected speeds along straight and curved pathways. For the first time for this mode of turning, plantar pressures were recorded using insole foot pressure sensors while participants' movements were simultaneously tracked using marker-based 3D motion capture. During the steady-state strides at the apex of the turn, the mean operating point of the inside ankle shifted by 1 degree towards dorsiflexion and that for the outside ankle shifted towards plantarflexion. The largest change in relative joint angle range was an increase in hip rotation in the inside leg (>60%). In addition, the inside foot was subject to a prolonged stance phase and a 10% increase in vertical force in the posteromedial section of the foot compared to straight-line walking. Most of the mechanical change required was therefore generated by the inside leg with hip rotation being a major driver of the gentle turn. This study provides new insight into healthy gait during gentle turns and may help us to understand the mechanics behind some forms of impairment.

Introduction

The locomotor system of a healthy human is energy efficient, robust to disturbances, and is able to perform versatile types of movement (Devine, 1985). Turning manoeuvres are estimated to comprise a substantial proportion of steps taken daily (Glaister et al., 2007). However, while straight-ahead walking and running have been studied extensively, fewer studies have focused on any kinetic, kinematic or foot pressure adjustments associated with turning (Courtine and Schieppati, 2004, 2003a, 2003b; Dixon et al., 2014; Glaister et al., 2008; Hase and Stein, 1999; Orendurff et al., 2006; Strike and Taylor, 2009; Taylor et al., 2005). Turning manoeuvres pose greater demands on the locomotor system relative to straight-ahead walking as the kinetics and kinematics of the bilateral limbs become asymmetric, causing greater mediolateral instability (Courtine and Schieppati, 2003b; Patla et al., 1999). The biomechanics of turning merits further study as previous research has highlighted that the elderly (Thigpen et al., 2000), individuals with Parkinson's disease (Guglielmetti et al., 2009), stroke patients (Godi et al., 2010) and transtibial amputees (Segal et al., 2011) struggle with turning in particular. Risks of falls and injuries are also greater during turning relative to straight ahead-walking (Cumming and Klineberg, 1994; Nevitt et al., 1991). A better understanding of how healthy people negotiate turns is important for identifying mechanisms underlying locomotor impairment.

Different types of turning manoeuvres are utilised during everyday activities (Glaister et al., 2007). Sharp turns, for example, are completed in just two to three steps, while more gentle turns, require several steady-state (turn continuation) steps in between turn initiation and termination steps (Glaister et al., 2007). A variety of turning manoeuvres have, therefore, been investigated in a laboratory setting. Some studies have identified and focused on the two main strategies used to navigate sharp turns, namely 'step' and 'spin' turns, which are defined by the change in direction being away from and towards the side of the limb in stance, respectively (Akram et al., 2010; Huxham et al., 2006; Patla et al., 1999; Taylor et al., 2005). Other studies have investigated steady-state turning manoeuvres such as walking circular pathways (Godi et al., 2014; Segal et al., 2008; Turcato et al., 2015); curved trajectories (Courtine and Schieppati, 2003a; Orendurff et al., 2006); and U-Turns (Guldemond et al., 2007; Hase and Stein, 1999).

The footfall kinematics, joint angle ranges and forces acting through the feet have not yet been quantified simultaneously in a single set of subjects for any given turning manoeuvre. Of the few studies that have investigated the modulation of ground reaction forces and impulses during turning, most of which have used force plates (Glaister et al., 2008; Strike and Taylor, 2009; Taylor et al., 2005). Force plates allow the complete analysis of the ground reaction forces (horizontal and vertical), however one drawback with force plate setups is that the participants may adjust their walking pattern in order to plant the desired foot on the force plate area (Glaister et al., 2008; Strike and Taylor, 2009). The surface area of the force plates also limits the number of steps that can be measured, meaning that the majority of research into foot pressure adjustments during turning focuses has focused on sharp turns completed in a few steps (Strike and Taylor, 2009; Taylor et al., 2005) and not gentle turns. One way to avoid this bias is to use insole pressure sensors as they do not influence the trajectory of the participant and allow data to be collected over a number of gait cycles. They allow the complete pressure map under the foot to be recorded but are unfortunately only capable of detecting normal forces (Turcato et al., 2015).

The purpose this study was to investigate any adjustments in vertical ground reaction forces and impulses during steady state gentle turning relative to straight-ahead walking in healthy subjects using insole foot pressure sensors for the first time. Additionally we sought to simultaneously quantify footfall kinematics and joint angle ranges using 3D motion capture for a more complete picture of gentle turning from a single sample of subjects. By using a gently curved pathway (radius ~2.75 m) we aimed to capture steady-state (turn continuation) strides at the apex of the curve for the inside and outside legs.

Methodology

Subjects

The protocol for this study was reviewed and approved by the University of Manchester Research Ethics Committee (reference number 13310). Informed consent was obtained from all participants. Twenty healthy subjects (10 male, 10 female), aged 19-30 years (mean age 23.6 ± 3.41 SD), mean body mass index 22.23 ± 2.3 SD, were recruited. All subjects, by self-report, were free from musculoskeletal injuries, disease or any other limitation that might alter natural locomotor patterns.

Procedure

The walking tasks were performed indoors, and included gentle left and right turns (LT and RT, respectively), and straight-line (SL) walking as the control. All paths (0.5 m wide) were marked out on the floor using tape. The path for the LT and RT tasks consisted of a 3 m straight-line segment leading up to a curved segment turning 90 degrees (radius of curvature 2.5 m for the inner, and 3 m for the outer tape) and another 3 m straight-line segment leading away from the bend (Figure 1A). LT and RT were performed on the same track by starting at either end of the track. The track for SL task consisted of a single 6 m straight segment (**Figure 1A**). A short straight-line path was preferred over a longer track or treadmill set-up in order to facilitate comparisons against similar short turning trajectories, and to more readily reflect sporadic short-burst ambulation within the home and community. All activities were conducted at the subjects' self-selected comfortable walking speed. Each walking task was repeated ten times and the order of the trials was randomised and rest periods given. Prior to data collection, a practice period was allocated for the subjects to familiarize themselves with the tasks.

Data Acquisition

Motion Capture

Spherical reflective markers were attached to the participant's skin with toupee-tape on both sides of the body at the ankles (lateral malleolus), knees (lateral epicondyle of the femur), hips (anterior superior iliac spine), the chest (suprasternal notch), and the shoulders (most lateral point of the acromion). The footwear was covered with a Velcro-friendly shoe wrap (Mocap Foot Wrap, Optitrack, Oregon, USA), and markers attached to the shoe wrap at the location of the heel, the tip of the toe, and at the 1st and 5th metatarsals. Three more markers were attached to the Velcro-friendly headwear (Mocap Beanie, Optitrack, Oregon, USA) at the centre of the forehead, above the right ear, and centre top of the head. The 3D position of the reflective markers were recorded at 120Hz using 6 or 8 motion capture (MoCap) cameras (Flex 13, Optitrack, Oregon, USA) with the software 'Motive' (v1.6.0 beta, Optitrack, Oregon, USA). Before the walking tasks, the marker positions were recorded whilst the participant was stationary in an upright position.

Foot Pressure

During all activities, plantar pressure was measured at a frequency of 100 Hz per sensor using the F-Scan VersaTek wireless insole system (Tekscan, Boston, USA) with a sensel resolution of 5.08mm × 5.08mm. Data was collected from both the left and right feet. Each insole was step calibrated according to the manufacturer's instructions prior to data collection. All subjects wore footwear of the same type (flat Converse-style canvas low top shoes with no arch support) corresponding to the individual's foot size. The recording of the pressure data was synced to the start of the MoCap recording using the Motive's sync output and VersaTek's sync input option.

Data Processing

Motion Capture Data

The markers recorded whilst the participant was stationary were manually labelled in the software 'Motive' and exported as an ASCII (.csv) file. All other recordings were exported unlabelled. A rigid-body skeletal model of an average human, which had previously been created for GaitSym (www.animalsimulation.org), was used as our standard model (see (Sellers et al., 2010) for further details). It consisted of a 3D mesh model of the human skeleton, ball joints at the hips, and single-axis hinge joints at the knee and ankle. The whole upper body (trunk, arms, and head) was kept as a single rigid body. Custom-written routines in the commercial software Matlab (MATLAB R2014a, The MathWorks Inc., Natick, MA, USA) were used to: 1. Label and trajectorise the marker data; 2. create a subject-specific scaled version of the skeletal model and match it to the marker data; and 3. extract joint angles, stride lengths and walking velocity. The matched trajectories and joint angles were filtered using a second-order low-pass Butterworth filter (applied in forward and backward time) with a cut-off frequency of 5Hz (Winter et al., 1974). The Matlab files used for processing are included in the supplementary information.

For both the left and right legs, data were extracted for an entire stride (the *apex stride*) at steady-state motion. The *apex strides* were selected from heel strike to heel strike (HS) at the apex of turning, which was defined as the time t_{45} at which the trunk completed 45 degrees of the 90 degree turn (see **Figure 1B** for further details). In the control experiment, i.e. walking straight, the *apex strides* were chosen accordingly around the time the trunk completed half the track length. The following quantities were extracted from the *apex strides*: stride length, average velocity and inclination of the trunk, and the minimum maximum and range of motion of the ankle, knee and hip joint angles.

Pressure Data

Frame data for each trial was exported from F-Scan Research software (Tekscan, version 7.19) as an ASCII file and imported into Matlab for further processing. Custom-written routines were used to: 1. identify heel-strike and toe-off timing; 2. ensure synchronisation with the MoCap data; 3. extract the following data during the *apex strides*: the mean vertical force, vertical force impulse, stride period, swing duration, stance duration and duty factor (time spent on stance leg as a fraction of the time of the entire stride). Additionally, the centre of pressure (CoP) was computed in a time window □t_{CoP} (rather than during the entire stance duration) between peak force at heel strike and toe off (see **Figure 1C**) because the sensel readings at initial heel contact and at forefoot push-off were too scattered to provide reliable data. Please refer to the supporting information for a more detailed description of the data processing.

Statistical Tests

The data of each participant (and each task) were averaged over the number of repeats. Commonly, the gait of healthy subjects is assumed to be symmetrical, however, several studies have found left-right asymmetries during walking tasks (Sadeghi et al., 2000). Hence, the analysis was performed separately for the left and right leg, using the data from the straight walking task as a control measurement. Differences between variables recorded during straight-line walking vs. those from curved trajectories were tested for significance via paired Student's t-tests in Microsoft Excel (2010).

Shapiro-Wilk tests were used to investigate whether the differences between curved and control task measurements followed a normal distribution, and a signed-rank test was used instead of paired t-test in cases where the differences were not normally distributed. Normality and Signed-Rank tests were performed with the Real Statistics Resource Pack Add-In (Release 4.10) for Excel (Zaiontz, 2016).

Results

Figure 2A shows the mean and standard error of mean of the velocity of all 20 participants. All participants decreased their walking speed for the turning stride by a statistically significant mean amount of 0.055 m/s ($p < 0.0001$ for both left and right turns). The stride length of the inside leg was shortened by a mean value of 0.064 m ($p < 0.0001$, **Figure 2B**) and the duty factor was increased on the inside leg and decreased on the outside leg (**Figure 2C**). Stride duration, however, was kept fairly constant during turning (**Figure 2D**). The increases and decreases in duty factor in the inside and outside legs respectively were achieved by modulation of both the swing and stance durations. For the inside legs, stance duration (**Figure 2E**) was increased and swing duration (**Figure 2F**) was decreased. For the outside legs, the opposite was true (**Figures 2E-F**).

Figure 3 shows the shift of the mean centre of pressure (CoP) in mediolateral and anteroposterior direction. **Figure 3A** depicts sample data of the CoP trajectories recorded on the left foot for repeated trials (of a single participant) and the associated mean CoP. **Figure 3B** shows the mediolateral change, and **Figure 3C** the anteroposterior change of the mean CoP during turning compared to walking straight. While the direction of the mean CoP shift in mediolateral direction was consistent between the left and right foot, only the data for the right outside foot was significant ($p < 0.01$). The CoP moved towards the posterior side (mean shift of 4.9 mm) in the inside foot, and towards the anterior side (mean shift of 4.1 mm) in the outside foot (**Figures 3C**).

Figure 4 contains tabulated values of the mean vertical force and the vertical impulse. When computing the mean vertical force and vertical impulse over the entire foot, significant differences were only found on the inside foot where the mean vertical force increased by 2.9% and the vertical impulse by 4.6%. Considering separate areas of the foot however revealed significant changes in the outside foot including increases in the anterolateral portions and decreases in the posteromedial sections of the foot. These differences must in effect, cancel each other out for whole foot measurements.

Figure 5 summarises the joint angle range during straight-walking and turning left and right for the left leg. Mean minimum and maximum joint angles and range of motion for each leg are presented in

Table 1. For the inside leg, ankle dorsiflexion increased while plantar flexion decreased; both knee and hip flexion increased while extension decreased; hip rotation (internal and external) increased and hip abduction decreased while adduction increased. By contrast, for the outside leg, the opposite was true for the angle of the ankle, the flexion and extension of the hip and the abduction and adduction of the hip. However, relative to straight-line walking there were no changes in the angles of the knees nor the rotation of the hips during turning in the outside leg.

Overall range of motion (RoM) differed significantly in fewer instances. In the inside leg, RoM increased for hip rotation and abduction/adduction. The largest difference was found in the hip rotation of the inside leg. Both internal and external rotation increased by an average of 4 to 5 degrees, leading to an average increase RoM by 9 degrees. This corresponds to a relative increase of RoM of 63% during turning compared to walking in a straight line. For the outside leg, RoM increased for the flexion/extension of the hips and the knees (right leg only). Additionally, the trunk was found to be leaning inwards (towards the curve) by 3 degrees and exhibit a larger range of swaying (**Table 2**).

Discussion

The participants were free to walk at their comfortable, self-selected walking speed during straight-line walking and turning. While turning, the participants decreased their walking speed relative to straight walking by shortening the stride length of the inside leg. Additionally, the stance phase of the inside leg and the swing phase of the outside were increased, both contributing to an increased duty factor for the inside and reduced duty factor of the outside leg. Our results agree with previous publications regarding the decrease in velocity for turning (Courtine and Schieppati, 2003b; Orendurff et al., 2006; Turcato et al., 2015) and the increase in stance time of the inside leg (Turcato et al., 2015). While the participants in this study maintained an almost constant walking frequency throughout all trials, the study by Turcato et al. (2015) revealed a significant decrease in stride frequency for turning, which may be due to their more challenging experimental setup (radius of curvature 1.2 m, participants completing three 360 turns).

During the stance phase, the CoP moves from the rear to the front of the foot in a curved trajectory from lateral to medial (Grundy et al., 1975). In the present study, the mean location of the CoP was computed and compared between walking straight and turning strides. The mean CoP moved towards the posteromedial area of the inside foot, and towards the anterolateral area of the outside foot during turning but the shift appears to be surprisingly small. This may partially be explained by the fact that the CoP was extracted only during mid-stance (see **Figure 1C**) because the data at heel strike and toe off were too noisy. As there are no previous studies available for comparison, further experiments are required to establish if and how the change of CoP is influenced by the type of turning manoeuvre (such as radius of curvature or walking speed).

Previous studies have found significant differences in the impulse in the mediolateral and anteroposterior direction using force plate data (Strike and Taylor, 2009; Taylor et al., 2005) but no detailed information was given on the impulse due to changes in the vertical force. When considering the entire foot area, a significant change in vertical force and impulse was only found for the right inside foot. Analysing different sections of the foot revealed however significant changes in both the inside and outside foot. The location of where the vertical forces acted on the outside foot clearly shifted from the posteromedial (relative decrease by approximately 13%) to the anterolateral (relative increase by approximately 9%) section during turning. This result agrees with the vertical force increase found in the II – V Metatarsal heads, and the force decrease in the medial heel reported by (Turcato et al. (2015) during circular walking. **Figure 4** also shows that the changes found in the inside foot were almost entirely due to an increase of vertical force (mean increase 10%) and vertical impulse (mean increase 12%) in the posteromedial section of the foot.

The majority of the joint angle ranges of the lower limb were modified by a significant amount to accommodate curved trajectories. In agreement with previous studies(Courtine and Schieppati, 2003b; Turcato et al., 2015), the participants leaned towards the turn (mean inclination of 3 degrees, $p <$ 0001). While the total RoM did not change for the ankle, the mean operating point shifted by 1 degree towards dorsiflexion for the inside, and towards plantarflexion for the outside ankle. Unlike previous studies (Taylor et al., 2005), our model used a single-axis joint at the ankle. Thus, any rotary ankle movements in the transverse plane would appear as hip rotation. The inside knee experienced an increase in maximum flexion ($p < 0.01$) during turning. This may be connected to the trunk inclination as the maximum knee flexion occurs during swing phase to achieve ground clearance (Kadaba et al., 1990).

The hip joint experienced the largest changes for turning versus walking straight. While the outside hip experienced a mean abduction increase of 2 degrees ($p < 0.0001$), the largest modifications occurred in the inside hip. The inside leg produced extra (external) rotation in order to place the leading foot at an angle to accommodate the curvature of the path. While the inside foot remained planted on the floor, the trunk rotated around the longitudinal axis (while the external foot was in swing phase) and surpassed the angle of the stance foot, resulting in an increase in the hip's internal rotation and adduction. Interestingly, this sequence of movements is similar to the 'spin turn' performed during sharp corners where the inside foot is placed into the corner apex, acting as a pivot for the trunk an outside leg to swing around. Previous research has shown that 'spin' turns occur regularly despite being less stable and more demanding (Akram et al., 2010). The fact that the inside leg is used as a pivot during gentle turns may be an indicator as to why 'spin' turns occur in more abrupt turning manoeuvres as well.

Conclusion

We investigated changes occurring in the locomotion pattern of healthy subjects during gentle turning compared to straight-line walking. In line with previous publications, participants decreased their walking speed during turning and increased the stance duration of the inside foot. Relative changes of around 10% were found in the vertical force and impulse in the anterolateral and posteromedial sections of the foot. An increase in maximum joint angles across the lower limb was found during turning compared to walking straight. The largest change was found in the hip rotation of the inside leg where the total range increased by over 60 % during turning compared to straight-line walking. It is the inside leg that was affected most by the adjustments made during locomotion along a curved trajectory. This included having to sustain longer stance phases, as well as a major increase in hip's (internal and external) rotation while the leg was loaded during stance phase. Our findings clearly indicate that it is the inside leg that acts as the main pivot in gentle turns.

Conflict of Interest Statement

The authors do not have to disclose any financial or personal relationships with other people or organizations that could inappropriately influence (bias) their work.

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Figures

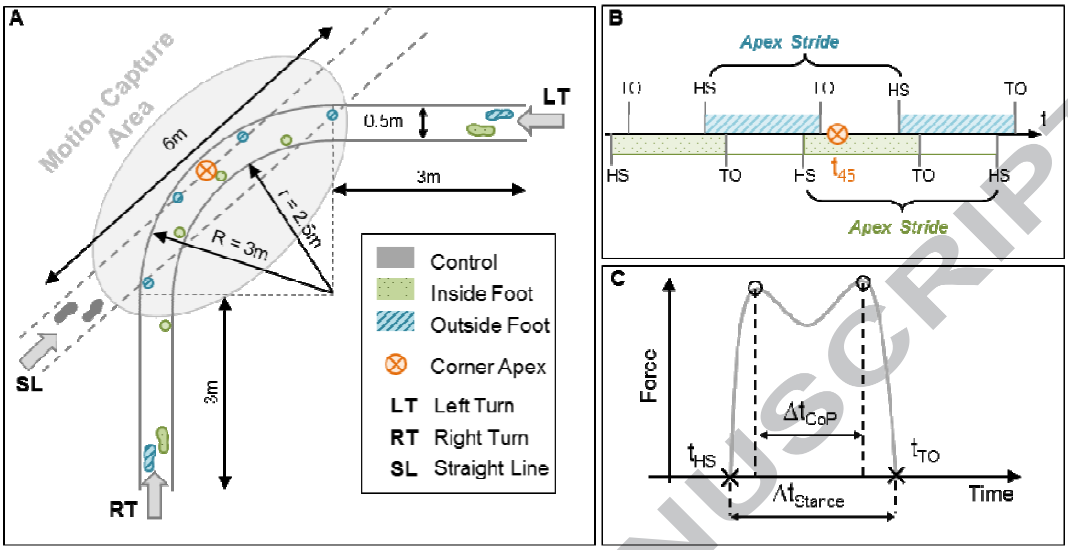

Figure 1: Pathways and the stride of interest. **A)** Separate pathways for walking straight and turning. Green and blue circles represent a typical striding pattern for a turning trial. **B)** Definition of the apex stride: The *apex strides* fulfil the condition: $t_{HS1} < t_{45} < t_{HS2}$, where t_{45} is the time the trunk reaches 45°, and t_{HS1} and t_{HS2} are the first and second heel strike, respectively. **C)** Definition of stance time Δt_{Stance} and the time period Δt_{CoP} during which the centre of pressure was computed.

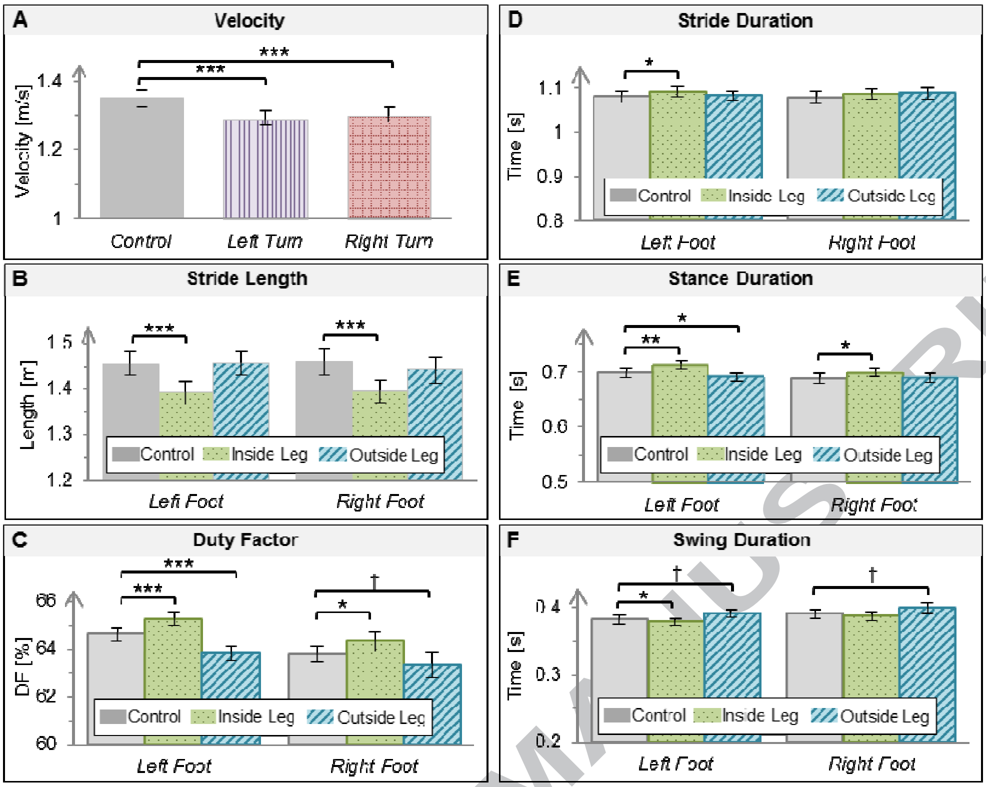

Figure 2: Velocity and stride parameters during straight-walking and turning. **A**) Mean velocity; **B**) Stride length; **C**) Duty factor; **D**) Stride duration; **E**) Stance duration and **F**) swing duration. Bar height represents average of all 20 participants and error bars represent ± 1 SE. Significance levels: * $p < 0.05$; ** $p < 0.01$; *** $p < 0.0001$, for data tested with a paired student's t-test and † $p < 0.05$ for non-normal data tested with a signed-rank test, between the control (grey) and inside (green, dotted) or outside (blue, diagonal lines) leg.

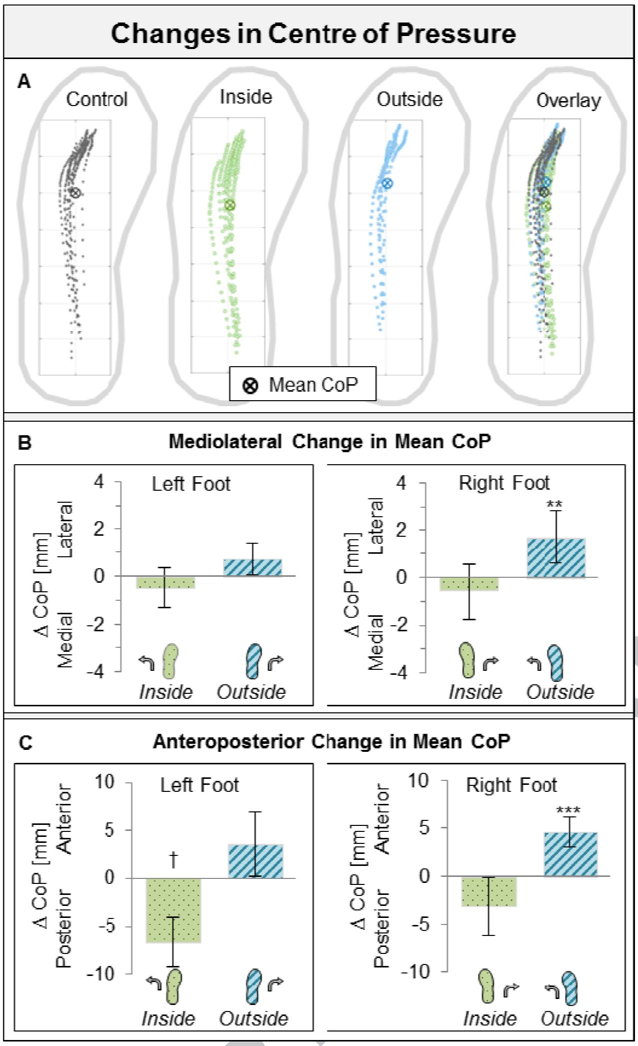

Figure 3: Shift of the mean CoP during turning versus walking straight. **A**) Sample data depicting the trajectory of the CoP during the time interval Δt_{CoP} (see **Figure 1C**). The data shows the tracks of one person for repeated trials. B) Shift of the mean CoP in mediolateral direction. C) Shift of the mean CoP in anteroposterior direction. Inset arrows (next to inset foot) indicate direction of turning. Error bars represent the 95% confidence interval. Significance levels: * $p < 0.05$; ** $p < 0.01$; *** $p < 0.0001$, for data tested with a paired student's t-test and † $p < 0.05$ for non-normal data tested with a signed-rank test.

	Entire Foot		Anterolateral		Anteromedial		Posterolateral		Posteromedial	
	Left	*Right*	*Left*	*Right*	*Left*	*Right*	*Left*	*Right*	*Left*	*Right*
Mean VerticalForce										
$F_{Control}$ [N]	543.2	538.4	163.3	157.4	155.1	149.6	123.8	134.4	100.9	97.0
F_{In} [N]	540.9	554.0	160.8	161.0	146.4	148.1	123.7	136.4	110.0	108.5
ΔF_{In} [%]	-0.4%	2.9% *	-1.6%	2.3%	-5.6% *	-1.0%	-0.1%	1.5%	9.0% **	11.9% ***
F_{Out} [N]	543.7	535.7	178.6	172.0	158.3	147.0	117.9	133.0	88.9	83.8
ΔF_{Out} [%]	0.1%	-0.5%	9.3% **	9.3% †	2.1%	-1.8%	-4.8% *	-1.1%	-11.9% †	-13.6% †
Vertical Impulse										
$J_{Control}$ [Ns]	385.7	375.5	115.8	109.7	110.1	104.1	88.0	93.9	71.8	67.9
J_{In} [Ns]	392.4	392.9	116.5	113.8	106.1	104.9	89.8	97.0	80.0	77.3
ΔJ_{In} [%]	1.7%	4.6% †	0.5%	3.8%	-3.6%	0.8%	2.1%	3.3%	11.5% **	13.8% ***
J_{Out} [Ns]	382.0	374.4	125.7	119.8	111.0	102.6	82.9	93.2	62.3	58.8
ΔJ_{Out} [%]	-1.0%	-0.3%	8.5% †	9.2% **	0.9%	-1.4%	-5.8% *	-0.7%	-13.2% †	-13.5% †

Figure 4: Mean vertical force and vertical impulse measurements during straight-ahead and turning *apex strides*. The differences between control (grey) and inside (green) or outside (blue) strides is given as a relative percentage value. Significant differences are highlighted with a darker background colour. Significance levels: * $p < 0.05$; ** $p < 0.01$; *** $p < 0.0001$, for data tested with a paired student's t-test, † $p < 0.05$ for non-normal data tested with a signed-rank test, between the control and inside or outside legs.

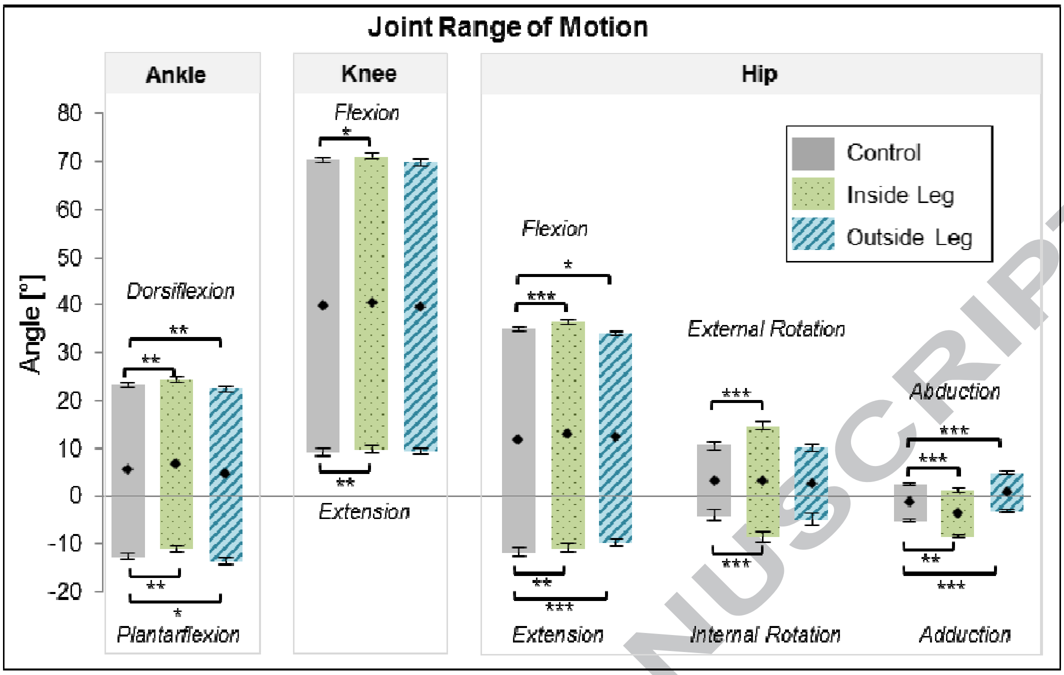

Figure 5: Joint range of motion of the left leg during *apex strides* while walking straight, turning left and turning right. The lower and upper ends of the bar show the mean minimum and maximum joint angles and the black diamond represents the mean of the range. Values for both legs are reported in **Table 1**. The error bars represent ± 1 SE. Significance levels: * $p < 0.05$; ** $p < 0.01$; *** $p < 0.0001$, for data tested with a paired student's t-test, † $p < 0.05$ for non-normal data tested with a signed-rank test, between the control (grey) and inside (green, dotted) or outside (blue, diagonal lines) leg.

Joint	Movement	Left Leg: Control (±1 SE)	Left Leg: Inside (±1 SE)	Left Leg: Outside (±1 SE)	Right Leg: Control (±1 SE)	Right Leg: Inside (±1 SE)	Right Leg: Outside (±1 SE)
Ankle	Dorsiflexion [°]	23.4 (±0.6)	24.4 ** (±0.5)	22.4 ** (±0.6)	23.0 (±0.7)	24.2 ** (±0.7)	22.0 ** (±0.8)
	Plantarflexion[°]	-12.6 (±0.7)	-11.0 ** (±0.7)	-13.5 * (±0.7)	-12.0 (±0.8)	-10.7 * (±0.8)	-13.7 *** (±0.9)
	RoM [°]	36.0 (±0.8)	35.4 (±0.8)	35.9 (±0.8)	35.0 (±1.0)	34.9 (±0.9)	35.7 (±0.9)
	Mean [°]	9.0 (±0.3)	10.0 ** (±0.4)	8.4 ** (±0.4)	9.2 (±0.4)	10.2 ** (±0.4)	8.3 *** (±0.5)
Knee	Flexion [°]	70.4 (±0.6)	71.2 ** (±0.6)	69.9 (±0.6)	70.2 (±0.5)	71.1 ** (±0.6)	69.1 *** (±0.5)
	Extension [°]	9.3 (±0.6)	9.8 * (±0.7)	9.4 (±0.6)	9.3 (±0.5)	10.1 ** (±0.6)	9.5 (±0.5)
	RoM [°]	61.1 (±0.6)	61.4 (±0.6)	60.5 (±0.6)	60.9 (±0.6)	61.0 (±0.6)	59.6 ** (±0.6)
	Mean [°]	30.8 (±0.5)	31.0 (±0.5)	31.0 (±0.5)	30.8 (±0.3)	31.6 *** (±0.4)	30.8 (±0.4)
Hip	Flexion [°]	35.0 (±0.4)	36.4 *** (±0.4)	34.1 * (±0.5)	35.0 (±0.5)	35.6 † (±0.5)	34.3 † (±0.5)
	Extension [°]	-11.7 (±0.9)	-10.8 ** (±0.9)	-9.7 *** (±0.8)	-11.6 (±0.8)	-10.5 *** (±0.8)	-8.4 *** (±0.7)
	RoM [°]	46.6 (±1.0)	47.3 (±0.9)	43.8 *** (±0.7)	46.6 (±1.0)	46.2 (±1.0)	42.7 *** (±0.7)
	Mean [°]	14.3 (±0.4)	15.3 *** (±0.5)	14.8 * (±0.5)	14.4 (±0.4)	15.4 *** (±0.4)	15.6 *** (±0.5)
	External Rotation [°]	10.5 (±0.8)	14.6 *** (±1.0)	10.1 (±0.8)	12.0 (±1.1)	16.1 *** (±1.2)	11.2 (±1.0)
	Internal Rotation [°]	-4.0 (±1.1)	-8.6 *** (±1.1)	-4.8 (±1.1)	-1.9 (±0.9)	-7.1 *** (±0.9)	-2.5 (±1.1)
	RoM [°]	14.6 (±1.0)	23.2 *** (±1.0)	15.0 (±0.9)	13.9 (±1.0)	23.2 *** (±1.2)	13.8 (±0.7)
	Mean [°]	2.5 (±0.8)	2.8 (±0.8)	2.5 (±0.8)	3.8 (±0.8)	4.0 (±0.9)	4.0 (±0.9)
	Abduction [°]	2.5 (±0.3)	1.2 ** (±0.4)	4.9 *** (±0.4)	2.8 (±0.4)	1.4 *** (±0.5)	4.5 *** (±0.3)
	Adduction [°]	-5.2 (±0.3)	-8.4 *** (±0.4)	-3.1 *** (±0.4)	-5.8 (±0.3)	-9.3 *** (±0.4)	-3.6 *** (±0.3)
	RoM [°]	7.7 (±0.3)	9.6 ** (±0.5)	8.0 (±0.6)	8.6 (±0.5)	10.7 *** (±0.7)	8.1 (±0.3)
	Mean [°]	-1.7 (±0.3)	-4.3 *** (±0.3)	0.5 *** (±0.3)	-2.2 (±0.2)	-4.7 *** (±0.2)	0.1 *** (±0.3)

Table 1: Maximum joint angle, mean joint operating point and range of motion. Significance levels: * $p < 0.05$; ** $p < 0.01$; *** $p < 0.0001$, for data tested with a paired student's t-test, † $p < 0.05$ for non-normal data tested with a signed-rank test, between the control and inside or outside leg.

Trunk	Control (± 1 SE)	Left Turn (± 1 SE)	Right Turn (± 1 SE)
Leaning Left [°]	-2.0 (±0.2)	-7.0 ** (±0.3)	0.0 *** (±0.3)
Leaning Right [°]	1.2 (±0.2)	-0.6 *** (±0.5)	5.6 *** (±0.5)
Range of Motion [°]	6.4 (±0.5)	7.8 *** (±0.7)	8.7 *** (±0.6)
Mean [°]	-0.4 (±0.2)	-3.4 *** (±0.4)	2.8 *** (±0.3)

Table 2: Trunk oscillation in mediolateral direction. Reported are maximum leaning angle, average range of motion and the location of the mean operating point. Significance levels: ** $p < 0.01$; *** $p < 0.0001$, for data tested with a paired student's t-test.