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Effect of walking speed on gait sub phase durations $\stackrel{\star}{\sim}$

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Spatiotemporal parameters

ABSTRACT

Gait phase durations are important spatiotemporal parameters in different contexts such as discrimination between healthy and pathological gait and monitoring of treatment outcomes after interventions. Although gait phases strongly depend on walking speed, the influence of different speeds has rarely been investigated in literature. In this work, we examined the durations of the stance sub phases and the swing phase for 12 different walking speeds ranging from 0.6 to 1.7 m/s in 21 healthy subjects using infrared cinematography and an instrumented treadmill. We separated the stance phase into loading response, mid stance, terminal stance and pre-swing phase and we performed regression modeling of all phase durations with speed to determine general trends. With an increasing speed of 0.1 m/s, stance duration decreased while swing duration increased by 0.3%. All distinct stance sub phases changed significantly with speed. These findings suggest the importance of including all distinct gait sub phases in spatiotemporal analyses, especially when different walking speeds are involved.

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50 1. Introduction

Spatiotemporal gait parameters are quantitative measures that describe gait performance (Hollman, McDade, & Petersen, 2011). They can be severely affected by musculoskeletal or neurological diseases (Cole, Silburn, Wood, Worringham, & Kerr, 2010; Leardini, O'Connor, & Giannini, 2014). Therefore, these parameters play an important role for classification between healthy subjects and patients at different stages of diseases showing pathological gait (Elbaz et al., 2014; Pradhan et al., 2015; Sen Köktaş, Yalabik, Yavuzer, & Duin, 2010) or for evaluating effects of interventions such as knee replacement surgeries in knee osteoarthritis (Levinger, Lai, Begg, Webster, & Feller, 2009; McClelland, Webster, & Feller, 2007). In elderly populations temporal variability and spatial parameters of gait can also discriminate between fallers and non-fallers (König, Taylor, Armbrecht, Dietzel, & Singh, 2014).

Basic temporal parameters usually include the duration of swing and stance phases, but distinct sub phases during stance should also be considered. Astephen and Deluzio (2005) determined a high discriminatory ability of the loading response

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61 phase in end-stage knee osteoarthritis patients. Stability of gait in elderly subjects might also be affected mostly during the loading response phase (Ihlen et al., 2012). As an additional factor, gait speed can serve to distinguish and compare healthy 62 and rehabilitating subjects based on spatiotemporal parameters (Andriacchi, Ogle, & Galante, 1977). The dependency of gait 63 phase parameters on gait speed might provide insights in understanding pathological gait as different responses to speed 64 have been observed in osteoarthritis and healthy populations (Bejek, Paróczai, Illyés, & Kiss, 2006). Further, gait speed 65 has been shown to be an important factor affecting the variability of temporal gait parameters (Kiss, 2010). Therefore, the 66 67 dependency of distinct gait sub phases on gait speed should be taken into account when analyzing gait using spatiotemporal 68 parameters.

A prerequisite in the use of temporal gait analysis is the quantitative definition of gait phases and their durations in healthy subjects. Perry (1992) stated a possible definition, splitting up the gait cycle into gait sub phases according to respective biomechanical tasks of each phase such as weight transfer or limb support (Fig. 1). In our study, we used this definition for separating distinct sub phases based on kinematic and kinetic data using an infrared cinematography system and an instrumented treadmill.

Only a few studies have investigated speed dependency of gait sub phase durations. Blanc, Balmer, Landis, and 74 Vingerhoets (1999) investigated stance, swing and double support times at self-selected speeds, but no investigation of 75 speed dependency was presented. Schwartz, Rozumalski, and Trost (2008) analyzed the effect of speed on different spa-76 tiotemporal, biomechanical, and neurophysiological parameters using kinematic and kinetic data. However, they roughly 77 78 defined five speed ranges by grouping self-selected speeds and investigated speed dependency of only double limb support, single limb support and swing phase. Liu et al. (2014) investigated gait sub phase variations of healthy subjects over different 79 self-selected velocities. A high-speed camera was used with manual gait event detection and 285 steps were analyzed in 80 order to describe the relationship between phase duration and speed. Although self-selected speed is a reasonable choice 81 in gait analysis, the speed definitions used are not applicable to accurately quantifying the effect of walking speed given vari-82 ability of the inter-individual perception of "normal", "slower than normal" and "faster than normal". Further, classification 83 84 into only three speeds, as presented by the authors, may result in loss of information as the speed ranges might overlap. A 85 comprehensive investigation of gait speed dependency of the duration of these sub phases over a wide speed range as well as modeling of this relationship using a large set of high resolution data is therefore still missing from the literature. We believe 86 that analyzing changes of gait sub phases with speed is a critical step towards a comprehensive spatiotemporal analysis of 87 88 gait.

The goals of our study were (1) to quantitatively describe healthy gait in terms of the proportional durations of sub phases during stance and (2) to model speed dependency of the durations by regression models.

91 2. Methods

92 2.1. Participants and preparation

Twenty-one healthy heel striking subjects (10 male, 11 female, age: 23.8 ± 3.3 years, height: 172.8 ± 9.4 cm, mass: 66.6 ± 10.9 kg) without injuries or musculoskeletal disorders participated in this study. The study was approved by the ethical committee of the University Hospital Erlangen (Re.-No. 106_13 B). All subjects gave informed consent before participating.



Fig. 1. Overview of the gait cycle and its sub phases that are analyzed in this study. Total gait cycle (HS–HS), stance (HS–TO, grey shaded), swing (TO–HS), loading response (HS–TO of contralateral leg), mid stance (TO of contralateral leg–HO), terminal stance (HO–HS of contralateral leg), pre-swing (HS of contralateral leg–TO). HS = heel strike, HO = heel off, TO = toe off. Definition of the phases according to Perry (1992).

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97 Six reflective markers were attached to each foot according to anatomical landmark definitions (Van Sint Jan, 2007) and 98 similar to a marker set used by Richards, Payne, Myatt, and Chohan (2014). They were placed on the shoes above the first, 99 second and fifth metatarsal heads, on the aspect of the Achilles tendon insertion into the calcaneus, as well as on the lateral 100 and medial malleoli as determined by palpation. The marker above the Achilles tendon insertion, which was not palpable 101 through the shoe, was placed on the same relative shoe position for every subject (Fig. 2).

102 2.2. Data acquisition

We collected three-dimensional kinematic data using a motion capture system (Qualisys AB, Gothenburg, Sweden) with 8 Oqus cameras sampling at 200 Hz. This system was set up around an instrumented split-belt treadmill (Bertec Corporation, Columbus, OH, USA) with integrated force plates sampling at 1000 Hz and with continuous synchronization between the two systems (Fig. 3).

Each subject walked at 12 different speeds (0.1 m/s increments starting at 0.6 m/s and ending at 1.7 m/s) after two min-107 utes of initial treadmill walking for familiarization with the equipment. Each subject was also allowed to acclimate to each 108 new speed setting for 30 s before the next measurement started. After familiarization, a 60 s long gait trial was recorded for 109 each speed setting. The chosen speed range covers very slow to very fast walking speeds (Bohannon & Williams Andrews, 110 2011) and also covers speeds of pathological or elderly gait. The speed order was unknown to the subjects and 111 pseudo-randomized, with a tendency to higher speeds presented at the end of the session to minimize fatigue effects. All 112 trials were performed continuously without stopping the treadmill. Before and after all measurements of each subject, 113 two seconds of data with unloaded force plates were recorded for subsequent force drift correction. 114

115 2.3. Data pre-processing

Marker coordinates and force data from the walking trials were filtered using a second order Butterworth low pass filter with a cut-off frequency of 6 Hz. We checked the data for steps spanning both belts at the same time. These steps were manually removed. Pre-processing was performed in Visual3D (C-Motion, Germantown, MD, USA).

119 2.4. Gait event detection

The following procedure was applied on the pre-processed data to calculate the sub phase durations: heel strike and toe off were detected for each foot separately based on a force threshold (20 N) in the vertical component of the ground reaction force. This threshold is high enough to avoid treadmill-induced noise at any of the tested speeds while maintaining a high detection sensitivity (Leitch, Stebbins, Paolini, & Zavatsky, 2011). In addition, a threshold of 0.1 m/s was applied to the component of the heel marker velocity perpendicular to the treadmill plane to detect the heel off event as previously described (Ghoussayni, Stevens, Durham, & Ewins, 2004). Using these events from both feet, the phases were subdivided into stance (loading response, mid stance, terminal stance, pre-swing) and swing phases (Fig. 1).

127 2.5. Statistical analysis and regression modeling

Further statistical analysis was performed in MATLAB (R2013b, MathWorks Inc., Natick, MA, USA) and SPSS Statistics (Version 22, IBM, Armonk, NY, USA). One gait cycle (from heel strike to heel strike) was normalized from 0% to 100% with sub phases being a proportion of this. For each walking trial and foot of each subject, mean and standard deviation of the phase durations were calculated.

A non-parametric statistical test with repeated measures (Friedman test) was performed to test for significant mean differences between all speeds for each phase separately after testing for normality (Shapiro–Wilk test) and sphericity (Mauchly's test). Between each sequential pair of speeds, a statistical post hoc test (Wilcoxon matched pairs signed rank test)



Fig. 2. Reflective foot marker setup in sagittal and frontal view.

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Fig. 3. Motion capture setup with instrumented treadmill.

was performed. Bonferroni correction was applied so that all effects are reported at a significance level of $\alpha = 0.05/66 = 0.00076$.

Sub phase duration modeling was performed using least-squares regression, fitting the mean durations of all subjects using the inverse of the variance as weighting factor, which accounts for unequal variances of the individual mean durations. We fitted three different models containing two estimated regression parameters (linear: f(x) = ax + b, inverse: f(x) = a/x + b, quadratic: $f(x) = ax^2 + b$) and we then calculated the coefficient of determination R^2 value for each model. The study data are available for download on www.activitynet.org.

142 **3. Results**

We manually labeled 615 steps spanning the two belts out of a total of 25,306 steps and we excluded them from further analysis. We visualized the individual means of the investigated phases using boxplots (Fig. 4).

Not all speed levels of the data fulfilled the normality assumption. Assumption of sphericity had been violated for all gait 145 phases. The Friedman test showed that gait phase durations changed significantly with speed for all phases (p < 0.05). 146 Statistically significant differences were found between most of the stepwise speed increments for each sub phase 147 (Fig. 4). The best generalization model was obtained with a linear fit for all sub phases (Table 1 and Fig. 4). Stance duration 148 149 decreased while swing duration increased by 0.3% per 0.1 m/s speed increment, respectively. Both loading response and 150 pre-swing phase decreased by 0.3%, mid stance phase decreased by 1.6% and terminal stance phase increased by 1.8% per 151 0.1 m/s step. The durations of the sub phases did not necessarily add up to the total duration of the stance phase, as labeling of the steps spanning two belts only excluded particular sub phases. 152

153 4. Discussion

In this study, we determined general relationships between gait sub phase durations and walking speed. The walking 154 155 speeds investigated here (ranging between 0.6 and 1.7 m/s) cover most of the expected everyday walking speeds of healthy 156 subjects (Bohannon & Williams Andrews, 2011). Walking speeds of elderly persons or patients are generally lower than those of young and healthy subjects, but they are still covered by the range studied – the mean speed of healthy elderly sub-157 jects older than 70 years has been shown to range from 0.98 to 1.17 m/s (Hollman et al., 2011). Patients moderately affected 158 by Parkinson's disease have been measured walking at an average speed of 0.94 (±0.21) m/s (Sofuwa et al., 2005), and the 159 mean speed for severely affected knee osteoarthritis patients has been shown to be $0.92 (\pm 0.24)$ m/s (Astephen, Deluzio, 160 Caldwell, & Dunbar, 2008). 161

All regression models and most changes in gait sub phase durations due to speed increments of 0.1 m/s were statistically 162 significant. The slope of each regression model quantifies the change of phase duration due to speed variations. The largest 163 effect observed was a redistribution of the mid stance phase towards the terminal stance phase with increased walking 164 speed. The variations of gait phase durations can be interpreted to some extent by biomechanical considerations. As walking 165 speed increases, stride length increases in normal gait (Stansfield, Hillman, Hazlewood, & Robb, 2006). Therefore, more time 166 167 is spent in the swing phase, necessary for leg progression, while the stance duration decreases (Tulchin, Orendurff, Adolfsen, & Karol, 2009). The double support phase, which consists of both pre-swing and loading response phases is known to 168 decrease with higher speeds (Stansfield et al., 2006; Tulchin et al., 2009). A shift of the heel strike of the contralateral foot 169 170 due to higher walking speeds might affect the sub phases of the ipsilateral foot (Fig. 1). Further information regarding the 171 shift of the heel off event is needed, however, to quantify this effect on the sub phases. Two potential mechanisms may

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Fig. 4. Boxplots of individual mean durations in percent of the total gait cycle vs. speed. The curves represent the linear regression lines. The results of Wilcoxon matched pairs signed rank tests with Bonferroni correction are shown for each adjacent speed pair (p < 0.00076).

Table 1

Results of the least squares regression of the gait sub phase durations. The estimated regression parameters are only given for the linear regression, which generalized best for all gait phases.

| | Linear | | | | Inverse | Quadratic |
|------------------|-------------|-------|----------------|---------|---------|-----------|
| Phase | a (%/(m/s)) | b (%) | R ² | р | R^2 | R^2 |
| Stance | -3.2 | 68.0 | 0.52 | < 0.001 | 0.53 | 0.49 |
| Swing | 3.4 | 31.7 | 0.56 | < 0.001 | 0.57 | 0.53 |
| Loading response | -3.2 | 17.9 | 0.56 | < 0.001 | 0.57 | 0.53 |
| Mid stance | -15.7 | 41.4 | 0.39 | < 0.001 | 0.32 | 0.40 |
| Terminal stance | 17.7 | -10.3 | 0.79 | < 0.001 | 0.76 | 0.78 |
| Pre-swing | -3.1 | 17.9 | 0.56 | <0.001 | 0.57 | 0.53 |

explain the shift from mid stance towards terminal stance phase: Firstly, higher speeds necessitate larger propulsive

impulses, and as the terminal stance phase is needed for force transmission and progression of the center of mass (Perry,

1992), more relative time needs to therefore be spent in the terminal stance phase. Secondly, as lower walking speeds lead to smaller stride lengths, the heel rise of the loaded foot may occur later during the stance phase, closer to the heel strike of

the contralateral foot. This results in a shorter terminal stance phase, as a result of the definition of the sub phases (Fig. 1).

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The calculated phase durations are in agreement with the speed independent durations determined by Perry (1992) for normal speeds around 1.3 m/s except for the mid and terminal stance phase. We calculated the mid stance phase to be longer and the terminal stance phase to be shorter than originally stated by Perry. This discrepancy can potentially be attributed to a different method of heel off event detection. The observations of decreasing stance and increasing swing phases are consistent with the literature. Andriacchi et al. (1977) reported a decrease of the absolute swing and stance durations, which was best estimated by quadratic polynomials. Schwartz et al. (2008) found that the stance phase and double support phase shortened with higher speeds, but the sub phases were not further evaluated. We found longer mid, and shorter terminal

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stance phases compared to Liu et al. (2014) for all speeds, while this systematic difference was smaller at fast speeds. In our study, this difference is equivalent to a later detection of the heel off event. This discrepancy may be due to the fact that: (1) our study was based on shod walking, while Liu et al. (2014) used barefoot walking; and (2) our event detection was based on a heel velocity threshold, while Liu et al. (2014) determined the event from video frames using the heel position. A more detailed quantitative comparison of speed related duration changes is not possible, as Liu et al. (2014) only reported the mean velocities for the three speed classes.

The high variances obtained in our study, especially of the mid and terminal stance phases, and the outliers reflect the 190 individual nature of the responses to different gait speeds. Much of the variation might arise from natural inter- and 191 192 intra-subject variability of gait as already described in the literature (Liu et al., 2014). Despite the statistical significance 193 of the models, the variability of the data lowers the coefficients of determination, and the responses to speed changes are 194 highly individualized, limiting the use of the models for precise predictions of gait phase durations of individual healthy sub-195 jects. Similarly, we cannot directly use this reference data to draw conclusions about a particular patient for clinical diagno-196 sis by quantifying deviations from our reference regression curves. However, disease-specific gait regression models could be used to quantify general differences between clinical and healthy populations. As the gait sub phases can be linked to biome-197 chanically meaningful tasks, potential deviations in the parameters between healthy and pathological models could be asso-198 ciated with the respective gait impairments. Therefore, similar experiments should be conducted in clinical populations to 199 obtain these disease-specific gait regression models. To this end, our dataset of healthy subjects may be used as a reference 200 and has therefore been made publicly available at www.activitynet.org. 201

The external validity might be compromised as treadmill-walking patterns may deviate from overground gait patterns. For most spatiotemporal parameters, only a few differences have been found between treadmill and overground walking (Lee & Hidler, 2008), but further evaluation should be performed for all gait phase durations. However, the main advantage of using a treadmill is the high number of steps that can be acquired.

Many different gait analysis setup ranging from gait mats (Elbaz et al., 2014) over ambulatory systems (Agostini, Balestra, Knaflitz, 2013; Pappas, Popovic, Keller, Dietz, & Morari, 2001; Rampp et al., 2014) to motion capturing setups can be used to extract gait phase parameters, which is an advantage over taking additional measurements. Therefore, similar analyses as presented here could be performed for different populations in common gait laboratories.

210 **5. Conclusion**

We quantitatively described the durations of the gait sub phase definitions using a system with a high spatial and temporal resolution. Gait sub phases were significantly dependent on gait speed and this dependency could be linearly modeled. The determination of all distinct gait sub phases can be potentially included in routine gait analysis. Future interventional and long term monitoring studies may reveal whether gait sub phase durations can be used to evaluate the effect of interventions, to monitor rehabilitation procedures and disease progression or as a discrimination criterion between disease stages. The investigation of gait sub phase changes with speed is potentially a further step towards a comprehensive analysis of gait.

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